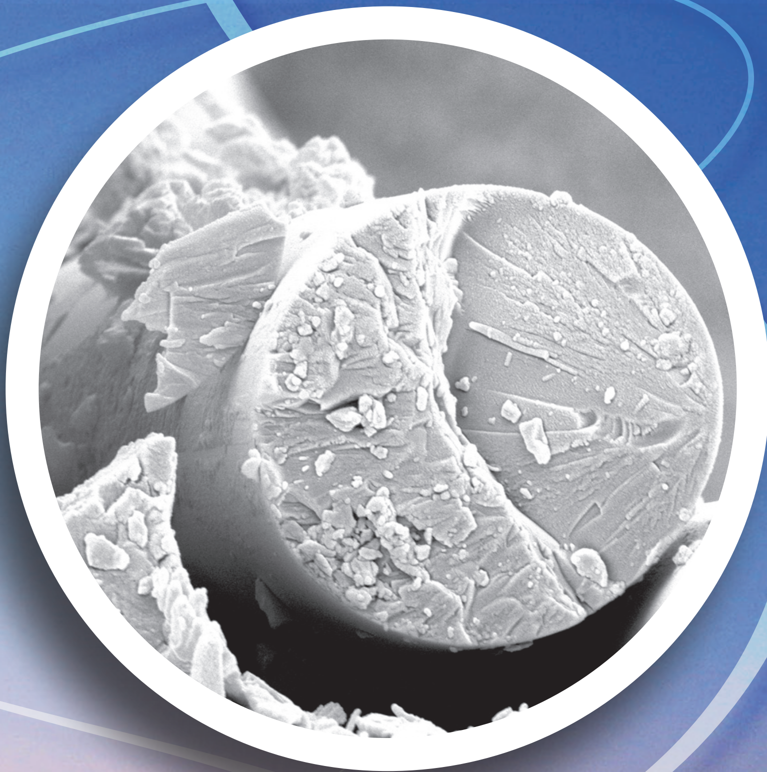

Exploring the limitations of fibre-reinforced composite fixed dental prostheses

Fibres (un)limited

Filip Keulemans

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Printed by: Drukkerij Michiels, Booischot, België.
Cover design by: Filip Van Reeth (www.nono.be).

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ACADEMISCH PROEFSCHRIFT

Ter verkrijging van de graad van doctor
aan de Universiteit van Amsterdam
op gezag van de Rector Magnificus
prof. dr. D.C. van den Boom
ten overstaan van een door het college voor promoties
ingestelde commissie,
in het openbaar te verdedigen in de Agnietenkapel
op donderdag 16 september 2010, te 14:00 uur

door

Filip Keulemans
geboren te Mechelen, België

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Faculteit der Tandheelkunde

“Do, or do not. There is no try”

Jedi Master Yoda

in Star Wars Episode V:The Empire Strikes Back

To Charlotte and Lise-Marie

This thesis was supported by IOT/ACTA and prepared at the Department of Dental Materials Science at the Academic Centre for Dentistry Amsterdam (ACTA), University of Amsterdam and VU University Amsterdam, the Netherlands.

CONTENTS

CHAPTER 1

Introduction

1.1 Introduction

A missing tooth is by many patients considered to be traumatic, especially when occurring in the visible region of the oral cavity. Common reasons for tooth loss are caries, periodontal disease, traumata and agenesis. Today, a dentist has a wide range of solutions available in case a single tooth is missing, *e*.g. orthodontics, autogenous tooth transplantation, removable dental prostheses (RDPs), fixed dental prostheses (FDPs) and implants. In spite of the cost effective and tooth preserving nature of autogenous tooth transplantation and RDPs their indication and use are limited [1,2]. Although orthodontic solutions have few indications on their own they are frequently used in combination with other prosthodontic treatment modalities. Prosthodontic solutions such as a conventional FDP, a resin-bonded fixed dental prosthesis (RB-FDPs) and a single-tooth implant are the most preferred options for the replacement of a single missing tooth and are acknowledged as the treatment of choice [3-5]. A single-tooth implant is mainly preferred when the adjacent teeth are intact and a sufficient amount of bone is available, while a conventional FDP is chosen in case the adjacent teeth are severely restored or the amount of bone is limited. On the other hand, a RB-FDPs can be an alternative treatment option in narrow single-tooth gaps neighbouring caries-free teeth [6]. Therefore, since conventional FDPs or single-tooth implants are not indicated for restoring all single-tooth gaps, the need for RB-FDPs is still persistent.

1.2 Resin-bonded fixed dental prostheses

Minimally invasive dentistry became the leading treatment strategy of contemporary dentistry [7]. The whole dental field, including restorative and prosthodontic dentistry adopted the concept of tooth tissue preservation. In the field of prosthodontics this paradigm shift can be noticed by the regained interest for RB-FDPs (Figure 1.1).

RB-FDPs have proven to be a reliable treatment alternative for the replacement of missing teeth [6]. A recent systematic review showed that RB-FDPs exhibit an estimated survival rate of 87.7% (95% confidence interval: 81.6%-91.9%) after 5 years [4]. Notwithstanding their good clinical performance, the most frequent complication was debonding, which occurred in 19.2% (95% CI: 13.8-26.3%) of RB-FDPs over an observation period of 5 years [4].

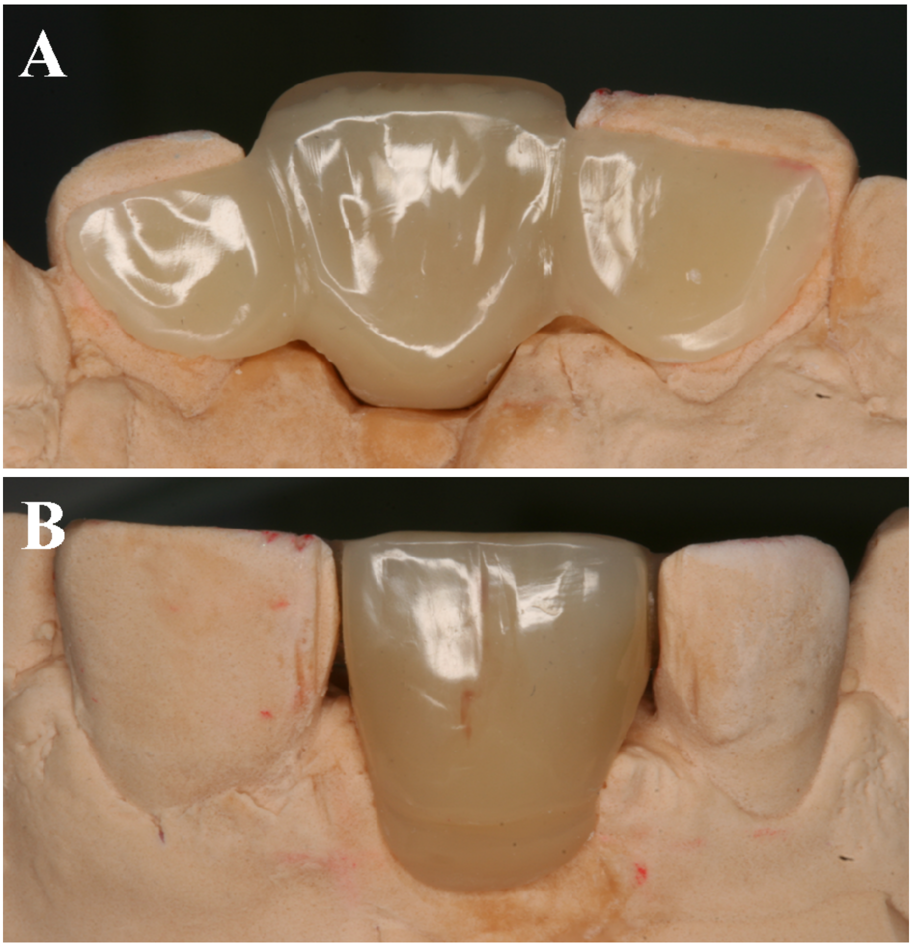

Figure 1.1 Three-unit fixed-fixed resin-bonded FRC-FDP: (A) Palatal view, (B) front view. (Courtesy of A. van Dalen)

Two strategies for decreasing the debonding rate of RB-FDPs are widely accepted. Several clinical studies stated that more extensive preparation of the abutment teeth, including palatal or lingual coverage with 180-degree wrap-around, chamfer, cingulum rests, and proximal guide planes and grooves, contributes to improve the retention of RB-FDPs [6,8-13]. Another way to minimize debonding is to design RB-FDPs as a two-unit cantilever. This approach came into focus after the observation that many partially debonded three-unit fixed-fixed RB-FDPs could be successfully converted into a two-unit cantilever design after removal of the debonded retainer [14]. Dynamic tooth contacts are believed to induce twisting and shear forces which cause retainers in fixed-fixed RB-FDPs to be dislodged; this is referred to as biting the tooth out of the retainer [8-10,15,16]. Elimination of interfacial stresses, because of their free-standing nature, provides a rationale for introducing two-unit cantilever RB-FDPs (Figure 1.2) in clinical practice [8,9,15]. Several clinical studies have demonstrated that two-unit cantilever RB-FDPs performed as well as or even better as their three-unit fixed-fixed counterparts [10,14,16-19].

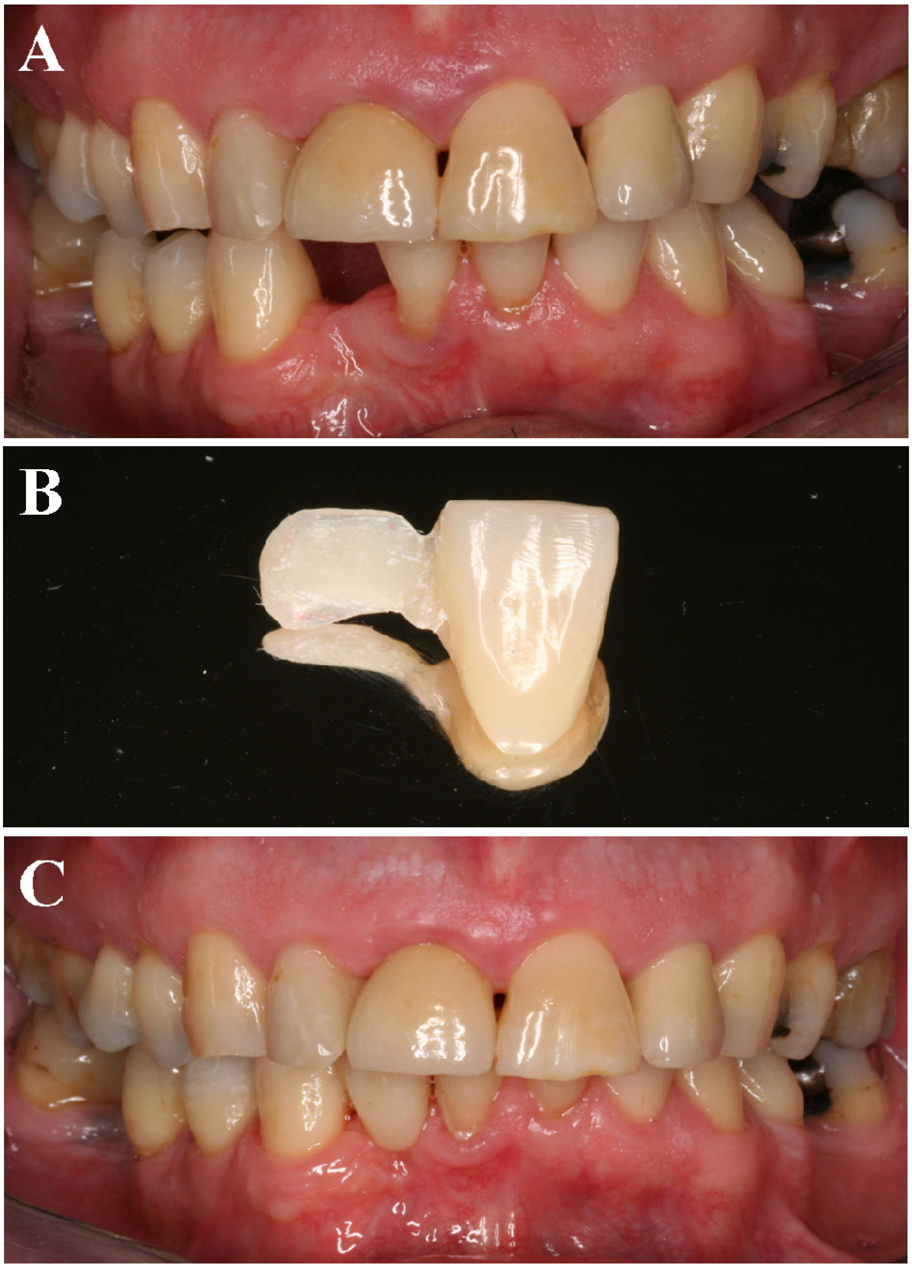

Figure 1.2 Two-unit cantilever resin-bonded FRC-FDP: (A) intra-oral view of the initial situation with a missing mandibular lateral incisor, (B) RB-FDP before cementation, and (C) intra oral view of the restored situation after cementation of the RB-FDP. (Courtesy of A. van Dalen)

The framework of RB-FDPs is traditionally made of metal alloys, but the undesirable greyish appearance of abutment teeth caused by shine-through of metal retainers in combination with the clinical reliability of two-unit cantilever RB-FDPs stimulated the interest in metal-free restorations, *e.g*. all-ceramics and fibre-reinforced composites. Nowadays, all-ceramics [19] and fibre-reinforced composites (FRC) [20] are viable alternatives for framework fabrication of RB-FDPs. Some clinical cases reported promising results for all-ceramic RB-FDPs [21,22]. In addition Kern *et al*. reported 5-year survival rates of 73.9 % for three-unit fixed-fixed designs and 92.3% for two-unit cantilever designs [19]. A recently published systematic review reported for FRC-FDPs a survival rate of 73.4% (95% CI: 69.4-77.4%) after 4.5 year [23]. During a 5 year multicenter clinical study FRC RB-FDPs exhibited a survival rate of 64% [24].

Since two-unit cantilever resin-bonded FRC-FDPs are already used for single tooth replacement in the anterior region [25-27], it is probably time to explore the viability of this treatment concept in the posterior region (Figure 1.3).

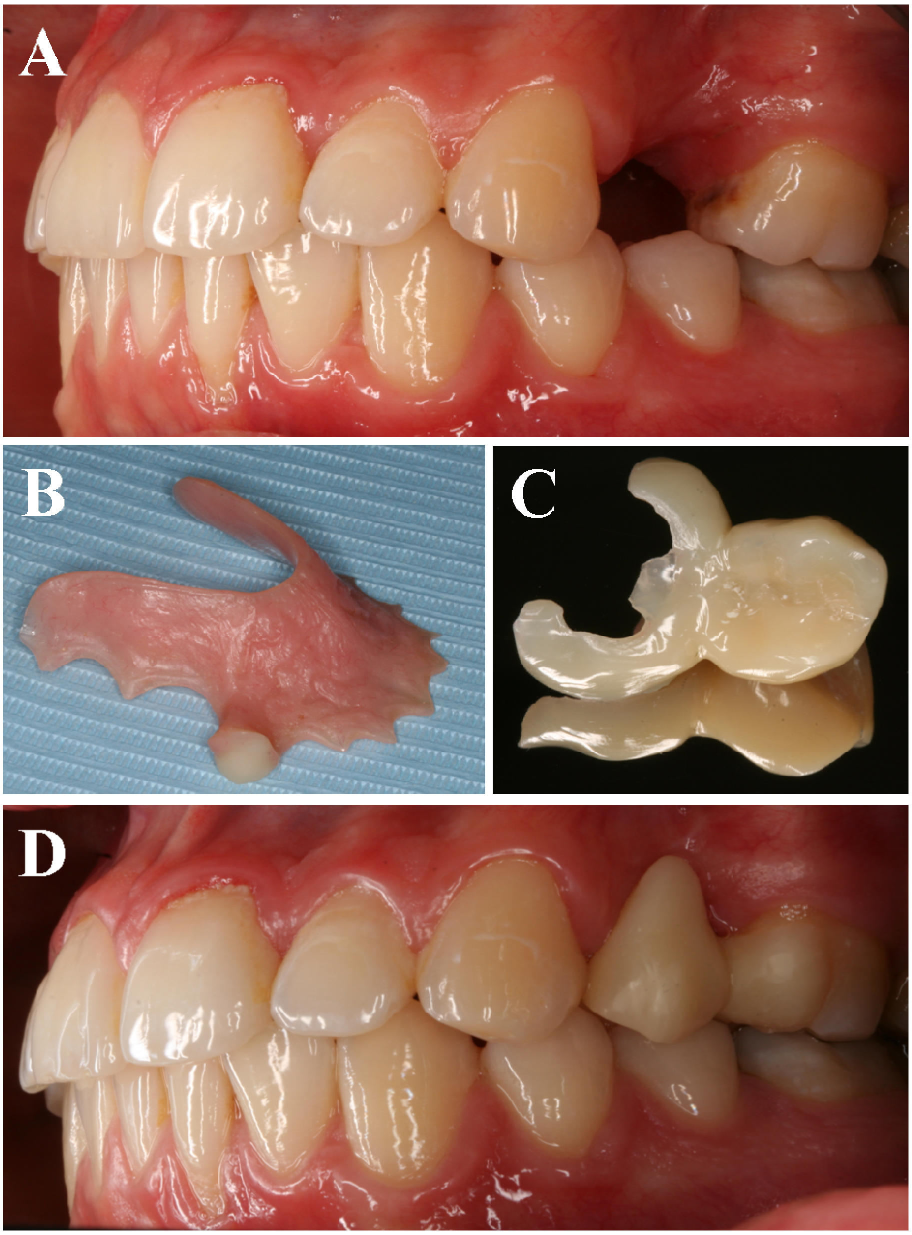

Figure 1.3 A patient with a missing upper first premolar (A) presented himself at the Oral Diagnostics Clinic of ACTA with a RDP (B). Due to the patients' new job, he requested a more comfortable fixed solution. Since he refused a conventional FDP and an implant because of financial implications, we provided him with a two-unit cantilever resin-bonded FRC-FDP (C and D).

1.3 Fibre-reinforced composite fixed dental prostheses

The use of glass fibres for the reinforcement of dental polymers was first proposed by Smith in the early 1960s [28]. Since the development and introduction of pre-impregnated glass fibre-reinforced composites in the early 1990s [29-36], they have been used in various dental fields like prosthodontics [37-42], implant dentistry [43,44], periodontics [45-48], paediatric dentistry [49-54], restorative dentistry [55-58], and orthodontics [59,60]. They are used in prosthodontic dentistry not only for the fabrication and repair of removable dental prostheses [61-63], but also for the fabrication of fixed dental prostheses, including crowns [64,65], bridges (Figure 1.4) [24,44,66], and resin-bonded bridges [67-70].

Several events contributed to the increasing popularity of FRC-FDPs. The first and most important event is the development of adhesive dentistry. The introduction of the acid etch technique by Buonocore in the mid 1950s [71] and the development of Bis-GMA as an organic matrix for resin composites by Bowen in the early 1960s [72] initiated the adhesive revolution and led towards the concept of tooth tissue preservation also known as minimal invasive dentistry [7,73]. Secondly, FRC-FDPs are restorations with a versatile fabrication procedure. They can be fabricated not only at the dental laboratory (indirect procedure), but also chairside (semi-direct procedure) or immediately into the mouth of the patient (direct procedure) by the dentist [44,65,66]. Thirdly, the community is becoming more aware of the possible adverse health effects of base alloys used in dentistry [74]. For particulate filler composite and glass fibre-reinforced composite, dental literature only provides suspicions of adverse health effects for patients [75] concerning the resin part, but no health effects of glass fibres are known so far. Last but not least, an increasing group of patients is seeking dental treatment for aesthetic enhancement. The amount of information regarding dental treatments that became available to our patients on the internet increased explosively during the last few years. Therefore, patients are aware of the aesthetic outcome of different kind of FDPs and prefer the excellent aesthetic properties of all-ceramic and the desirable aesthetic appearance of resin composites above the possible unacceptable aesthetic appearance of metal-containing FDPs.

Framework design of FRC-FDP.

The framework of FRC-FDPs is traditionally made of one or more bundles of unidirectional FRC, which span the entire length of the tooth replacement. Those fibre bundles are placed from the mesial abutment towards the distal abutment and slightly curved towards the gingival part of the pontic, in order to obtain a framework where the fibres are located in the area of the construction with the highest tensile stress, when the pontic is loaded, and the fibre direction is perpendicular to the occlusal load (Figure 1.5) [23,76-80].

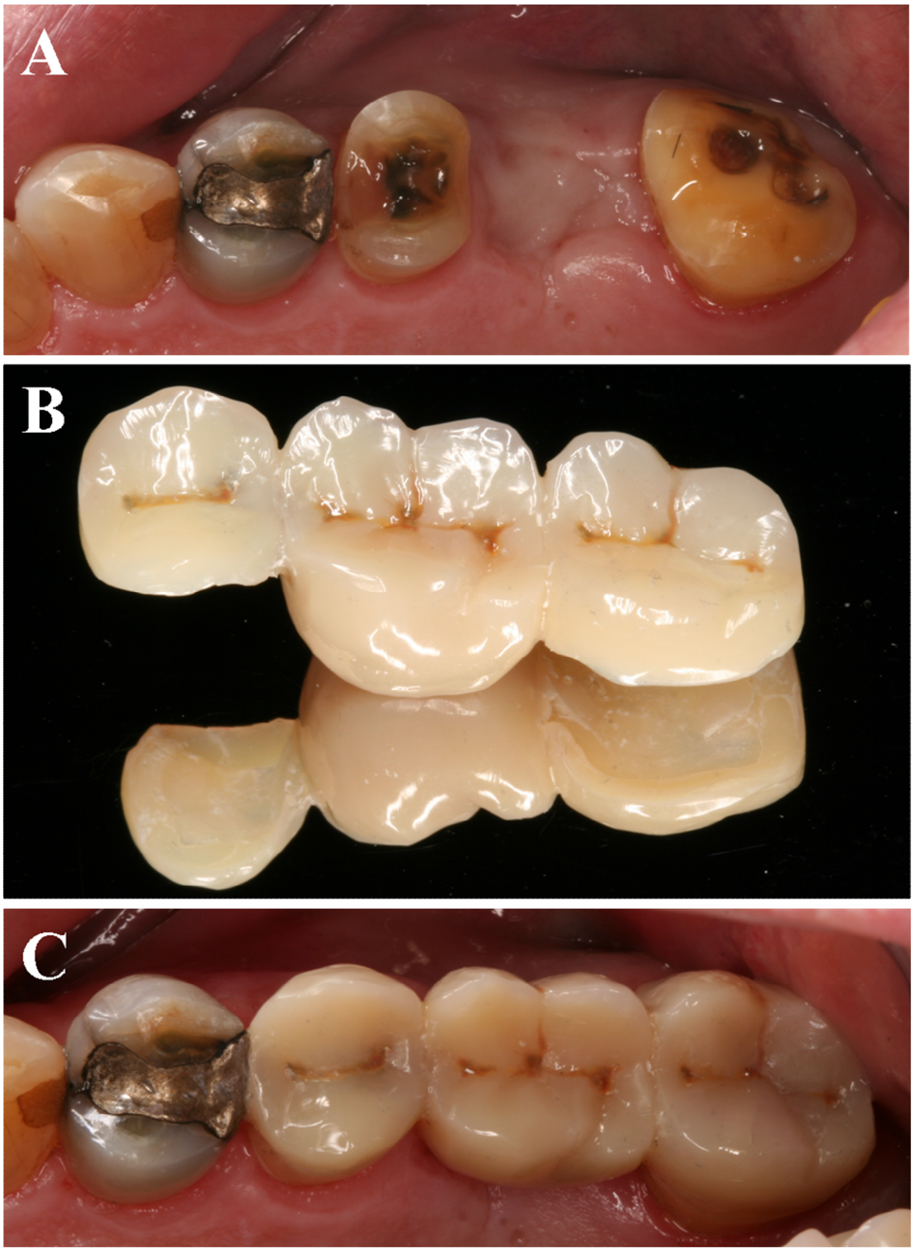

Figure 1.4 Three-unit onlay-retained FRC-FDPs replacing a missing molar in the upper jaw: (A) Onlay preparations on teeth 25 and 27, (B) FRC-FDPs before cementation, and (C) intra oral view of FRC-FDPs after cementation.

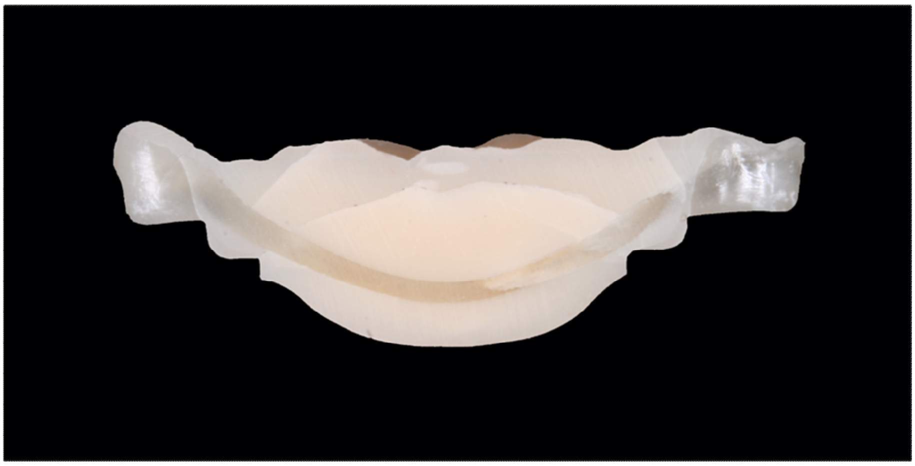

Figure 1.5 Inlay-retained three-unit FRC-FDP: cross-sectional view revealing the fibre location throughout the FDP.

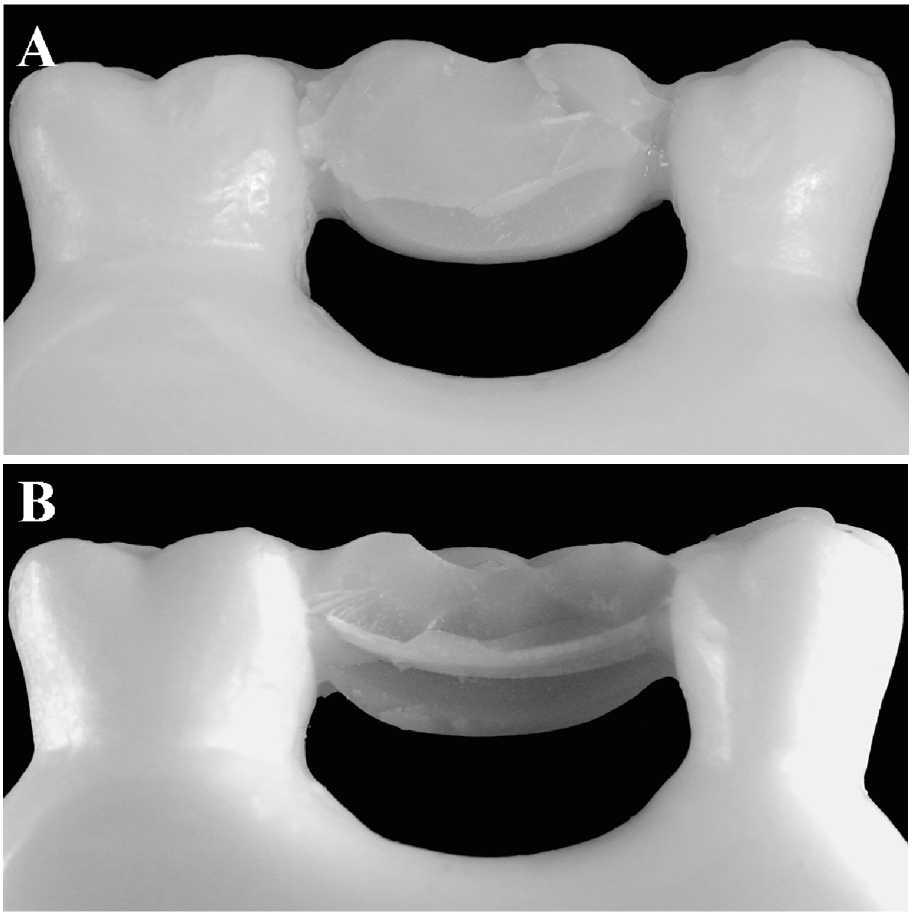

Figure 1.6 Clinical failures frequently encountered with FRC-FDPs: (A) chipping and (B) delamination of the veneering composite.

The most frequently encountered clinical problems with FRC-FDPs are chipping and delamination of the veneering composite [76]. To avoid confusion both terms should be explained. Chipping is a cohesive failure within the veneering composite (Figure 1.6A), while delamination involves adhesive failure between FRC framework and veneering composite (Figure 1.6B). The later results in exposure of the FRC framework. It is believed that the explanation for this problem lies in the design of the framework. The conventional framework design is lacking support to the veneering composite and rigidity of the FDP [76]. Several modified framework designs were proposed to overcome these shortcomings.

Freilich *et al*. [76] proposed a high-volume framework with increased rigidity and broader support for the veneering composite. A high-volume framework is fabricated by adding an amount of FRC to the pontic part of a traditional FRC framework and to enwrap it with a fibre bundle in such a way that the outer layer of fibres is perpendicular to the inner framework. A clinical study with a mean observation time of 3.75 years indicated a superior survival rate of 95% for the high-volume framework in comparison to a survival rate of 62% for the low-volume framework [79].

An anatomically-shaped framework was proposed by Monaco *et al*. [79] and Behr *et al*. [81]. Such an anatomically-shaped FRC framework supports the veneering composite completely by using extensions on the buccal and lingual side similar to the metal framework of a metal-ceramic FDP. The viability of such an anatomically-shaped framework was evaluated in an *in vitro* study, which compared the fracture resistance of simulated three-unit FRC-FDPs with different frameworks. The anatomically-shaped framework exhibited a fracture resistance of 902N which was significant higher than the 694N and 737N exhibited by conventional frameworks [81]. Monaco *et al*. [79] compared the clinical behaviour of inlay-retained FRC-FDPs with an anatomically-shaped framework to those with a conventional framework and found a not significantly different higher failure rate for the latter. Only 5% of FDPs with an anatomically-shaped framework suffered from chipping, while 16 % of FDPs with conventional framework suffered from delamination after an evaluation period of 12 to 48 months.

Xie *et al*. [82] proposed a framework which supports the pontic area in buccolingual direction by placing a short fibre bundle just under the occlusal surface of the pontic in a 90° angle to the direction of fibres of the main fibre framework. The proposed framework was compared to conventional and high-volume frameworks by evaluating the fracture resistance of inlay-retained FRC-FDPs. It was concluded that the framework with buccolingual support outperformed conventional as well as high-volume frameworks.

Garoushi *et al.* [83] introduced recently a semi-interpenetrating polymer network based random-orientated short fibre containing FRC (S-FRC) and explored its possible use for the construction of short-span FRC-FDPs. An *in vitro* study revealed that short-span FRC-FDPs made of S-FRC exhibited a comparable load-bearing capacity as FRC-FDPs with a traditional framework [84].

Clinical performance of fibre-reinforced composite fixed dental prostheses.

One of the major concerns of FRC-FDPs is that there is less evidence about the survival rate compared to metal, porcelain fused to metal, and all-ceramic restorations. Several clinical studies reported on the clinical performance of FRC-FDPs since the first clinical report was published in the early 1990s by Altieri *et al.* [64]. This first report described the clinical evaluation of three-unit FDPs constructed of an acrylic denture tooth dummy with a framework of an experimental glass fibre/polycarbonate matrix FRC. The FDPs were bonded to unprepared abutment teeth by bucally and lingually placed wing-shaped retainers. Although limited success was encountered, with reported survival probability of 50% at 1 year, the potential advantages of FRC-FDPs were acknowledged.

Culy and Tyas [25] concluded after 10 months that directly made cantilever FRC-FDPs could be a viable treatment option for the replacement of missing teeth in the anterior region.

A study by Freilich *et al.* [76] evaluated 39 three-unit FRC-FDPs in 25 patients of which 22 FDPs had extracoronal retainers and 17 FDPs had intracoronal retainers. The FRC framework was fabricated of pre-impregnated unidirectional FRC (FibreKor, Pentron, USA) The framework design also evolved during the course of the study. The original low-volume design was modified into a high-volume design (for more information see section 3.1) after the observation of delamination of the veneering composite only after three months. An overall survival rate of 74% was found after a mean service life of 3.8 years. A significantly higher survival rate was found for high-volume frameworks in comparison to their low-volume counterparts, 95% and 62% respectively. Although there was a trend towards higher survival rates for extracoronal retainers, no significant differences were found regarding retainer design.

Vallittu *et al.* evaluated a group of thirty-seven patients, who received resin-bonded FRC-FDPs, after 24 months [85] and after 42 months [20] of service. Different retainers, including wings, inlays and full coverage crowns were employed. The FRC frameworks were made of continuous unidirectional (Stick, Stick Tech Ltd, Finland) or woven bi-directional E-glass fibre-reinforced composite (StickNet, Stick Tech Ltd, Finland), which needed further manual impregnation with light-curing resins. Thirty-one patients were evaluated after 24 months and exhibited a survival probability of 93% .[85] The survival probability dropped towards 75% after 42 months of service [20]. The main reasons of failure were debonding and framework fracture.

Recently, van Heumen *et al*. [86] reported the results of a multicentre study on the long-term clinical performance of three-unit FRC-FDPs. Sixty anterior FRC-FDPs with a FRC framework made of Stick (Stick Tech Ltd, Finland) were inserted in fifty-two patients. Forty-eight FDPs were surface-retained and only twelve were hybrid-retained. An overall survival probability of 64% at five years was found. Once again, delamination of the veneering composite was the most frequently encountered problem seen in 47% of the cases, but did not always resulted in loss of the FDP. Most restorations were lost due to fracture of the connector area or a combination of problems.

Several studies evaluated the clinical performance of Targis/Vectris system (Ivoclar-Vivadent, Liechtenstein) for fabrication of FRC-FDPs.

In contrast to other studies investigating the short-term clinical survival of FRC-FDPs made of Targis/Vectris the study by Bohlsen and Kern found a rather low survival rate of 65.1% at three years [87]. In total, eighty-three FRC-FDPs were provided to thirty-nine patients. One should keep in mind that FRC-FDPs should preferably be luted with adhesive luting cement, which was not the case in this study. Twenty-two FDPs were cemented with a temporary cement and fifty-five were cemented with a zinc-phosphate or glass-ionomer cement. It was concluded that FRC-FDPs exhibited a lower survival rate than metal-ceramic FDPs and therefore could not be recommended as permanent restorations.

One short-term study compared two different metal-free restorative systems, namely one all-ceramic and one FRC system, and evaluated their potential for the fabrication of short-span inlay-retained FDPs [88]. Twelve FRC-FDPs made of Targis/Vectris (Ivoclar-Vivadent, Liechtenstein) were evaluated after a mean observation time of 15.3 months without any failures. The authors concluded that this type of restoration provided excellent aesthetics and saved tooth tissue.

Behr *et al*. [78] inserted twenty-two three-unit inlay-retained FRC-FDPs made of Targis/Vectris of which seventeen were adhesively fixed and five were conventionally cemented. The estimated cumulative survival rate was 72% at thirty-six months. Frequently encountered problems were chipping and delamination of the veneering composite, wear and discoloration.

Monaco *et al*. [79] inserted forty-one three-unit inlay-retained FRC-FDPs made of Targis/Vectris in thirty patients. A conventional framework was used in nineteen FDPs, while twenty-two FDPs received an anatomically-shaped framework. Delaminations of the veneering composite within one year of service were experienced with the conventional framework design in contrast to chipping of the veneering composite after 46 months of service for the anatomically-shaped design. Differences in failure rates between both designs are described in section 3.1. This study found an overall survival rate of 86% for FRC-FDPs and concluded that those restorations showed good clinical service over a short-term observation period (12 to 48 months). The same authors evaluated in a second study three-unit inlay-retained FRC-FDPs made of SR Adoro/Vectris luted with two different bonding systems, a two-step etch-and-rinse system (Excite DSC, Ivoclar-Vivadent) and a three-step etch-and-rinse system (Syntac, Ivoclar-Vivadent), respectively [89]. After a short observation time of 24 months a survival rate of 89.4% was observed for the two-step bonding group and 100% for the three-step bonding system. No post operative sensitivity during the first six months of service, nor debonding of the restorations were observed with the three-step bonding system.

Göhring *et al*. reported one year [90], two year [91] and five year [77] results of an ongoing long-term clinical study on posterior inlay-retained FRC-FDPs made of Targis/Vectris. Thirty-six patients received fifty-three FRC-FDPs. In order to reveal possible influence of predictors, such as age and gender, only one restoration in each patient was evaluated. Two cumulative survival rates were reported. The cumulative survival rate at five years was 97% for not debonding and 73% for not delaminating. Delamination of the veneering composite was reported as the main reason for failure. Other commonly observed non-catastrophic failures were chipping, occlusal wear, surface roughening and staining. Ninety percent of the margins were rated as perfect after 5 years. Significant changes regarding marginal adaptation were observed after one year of service, hereafter no further deterioration was seen. The authors concluded that FRC-FDPs performed acceptable, but that improvement of framework designs and materials was needed to prevent delamination.

Two systematic reviews were published in order to obtain a general idea on the clinical performance of FRC-FDPs. The first review was published in 2005 by Jokstad *et al*. [92]. It was concluded that insufficient evidence was available for advocating FRP-FDPs as an alternative to conventional FDPs and therefore be regarded as experimental. The authors emphasized that well-designed long-term clinical studies are necessary to point out the potential of FRC-FDPs as a permanent restoration. Recently, van Heumen *et al*. [23,93] published a systematic review and obtained an overall survival rate. Fifteen studies were included, which resulted in 435 FRC-FDPs obtained from 13 patient sets. The observation periods ranged from 10 months towards 5.7 years. They calculated an overall survival rate of 73.4% (69.4-77.4%) at 4.5 years. The most frequently reported failures were delamination of the veneering composite and fracture of the restoration. Once again, the need for well-designed randomized clinical trials was stressed .

1.4 Fibre-reinforced composites

Introduction

A composite is a combination of two or more materials which differ in shape and composition. When combined, these materials do not merge completely and they do not dissolve or react creating a new chemical substance. Each of the composite components maintains its properties and they are linked together with an interface. When used as a composite the properties of the materials improve beyond the level of their solitary use [94].

Composites are omnipresent in the world surrounding us, both as natural and man-made or as synthetic composites. Examples of natural composites are wood, containing cellulose fibres embedded into lignin and hemicellulose, bone, enamel and dentine, composed of hydroxyapatite (inorganic filler) and collagen and proteins (organic matrix). The observation that those natural composites exhibit high strength and low weight, made humans reflect on the possibility to design composites for different purposes.

One of the first human attempts to reinforce materials with fibres would probably be the incorporation of straw into walls made of clay as early as 4000 years ago. Iron-reinforced concrete is an industrial example of a composite, first used by the French building contractor François Coignet in 1853 for constructing a house at 72 Rue Charles-Michels in Paris.

These days when using the word “composite” this very often refers to the use of polymers. These polymers find their applications in every corner of society. Mid 20th century polymer production expanded to a worldwide scale. Fibre reinforced polymers have been part of our world ever since. In industry there are three important categories of fibres used in conjunction with polymers, i.e. glass, carbon and aramid. Each of these fibre categories are widely used for applications requiring polymers, or plastics as they are usually referred to, with specified properties like strength or elasticity. Well known applications are in the automobile industry, in aerospace, and sporting goods like tennis rackets, golf clubs and fishing rods.

Composition

In general, reinforced polymers or resin-based composites usually consist of inorganic fillers embedded in an organic polymer matrix. Normally the filler acts as a reinforcing material, while the polymer matrix binds all constituents together. Contemporary resin-based composites can be classified, depending on the size and volume fraction of the reinforcing phase, into three common groups [94,95]: (1) particle-strengthened composite or particulate filler composite (PFC) (Figure 1.7A), (2) fibre-reinforced composite (FRC) (Figure 1.7B) and (3) structural composite.

The filler particles used in PFC have an aspect ratio or length-to-diameter ratio close to 1, meaning that their dimensions are nearly the same in all directions. The fibre filler used in FRC has a much larger length-to-diameter ratio and can be in discontinuous and continuous form. The latter exhibiting a length-to-diameter ratio of up to 200. Structural composites are combinations of composites and homogeneous materials. An example of a structural composite is a laminate, which is composed of two or more layers of different materials that are bonded together, where each layer may contain fibres with different alignment.

One should be aware of the fact that the function of each component can differ according to the classification of the composite group [95]. In the case of PFC a distinction should be made between large particle strengthened composites and dispersion strengthened composites. In large particle strengthened composites the load is more or less evenly distributed between matrix and particles. This behaviour does not apply to dispersion strengthened composites, where the matrix bears the main load and the particles primary function is to obstruct the pathway of the cracks. From this it can be concluded that PFC are exhibiting high resistance to compressive stresses, meaning they are especially suited to function in supported load bearing applications. Dispersion strengthened composites are well established in the field of dentistry where they are extensively used for restoring cavities. In FRC the fibres bear the main load, while the primary function of the matrix is to hold the fibres together, to distribute the load and subsequently transfer it towards the fibres. Unlike PFC, FRC are especially resistant to tensile stresses, making them suitable in unsupported load bearing applications. An example is steel reinforced concrete used for the construction of floor systems in the building industry and unidirectional glass fibre-reinforced composite used for the fabrication of fixed dental prostheses in dentistry.

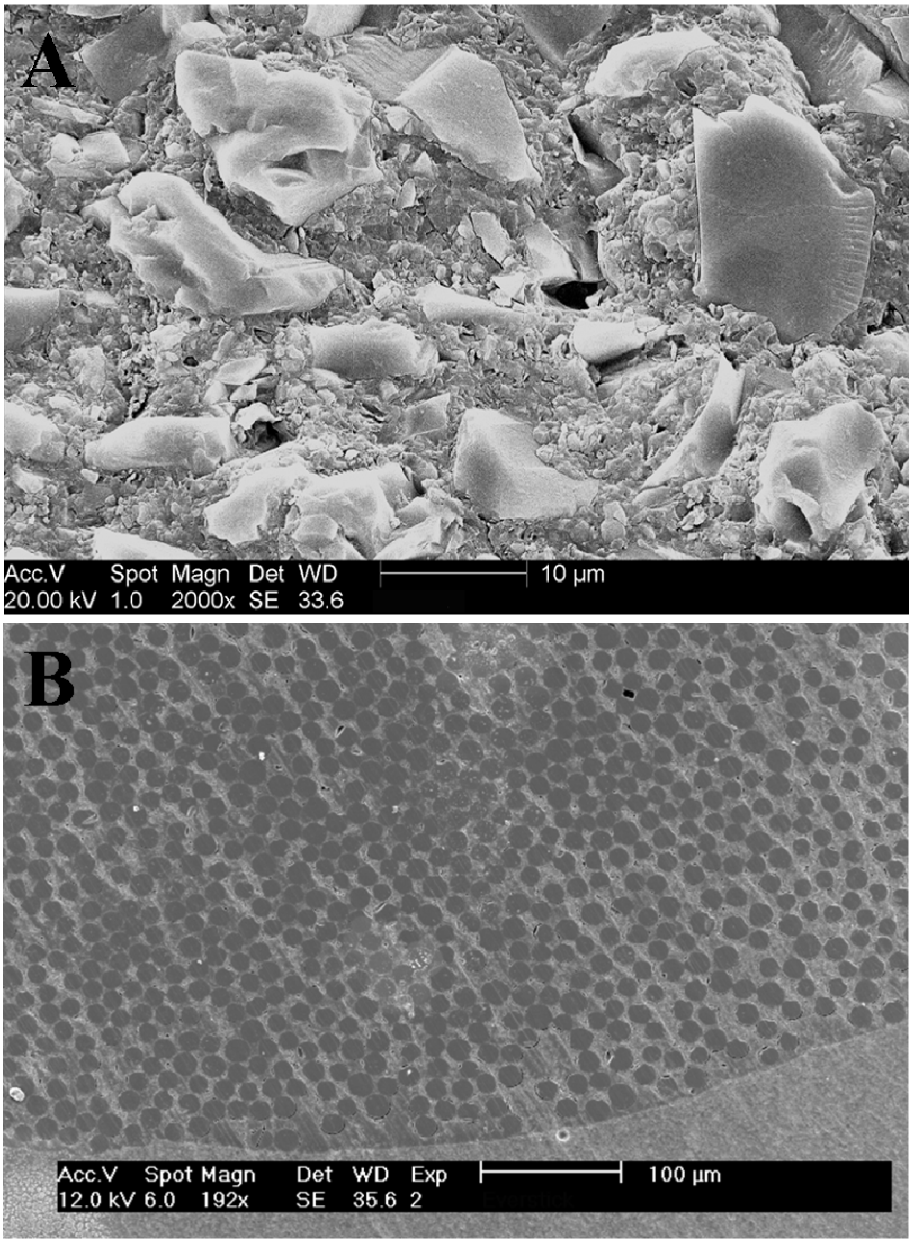

Figure 1.7 Dental resin-based composites: (A) particulate filler composite, and (B) fibre-reinforced composite.

The previous part intended to produce a brief overview on the classification and general composition of resin-based composites, while the next sections focus on the major constituents of dental fibre-reinforced polymers.

Matrix

As mentioned previously, the matrix forms a continuous phase wherein all constituents are embedded. Several functions can be ascribed to a composite matrix [95]: the matrix (1) binds components together and therefore preserves the desired fibre orientation and spacing, (2) protects the reinforcements from environmental influences, (3) distributes the applied load and subsequently transfers stress to the reinforcements, (4) provides durability and toughness, and (5) determines the thermo-mechanical characteristics.

In general, the FRC matrix encompasses around 30-40 % of the volume of the entire composite structure. The elastic modulus of the matrix is low in contrast to that of the fillers [95]. Well established dental polymers such as Poly(methyl-methacrylate) or PMMA and Bisphenol-A-glycidyl dimethacrylate or Bis-GMA exhibit an elastic modulus of 2 GPa [96] and 9 GPa [97], respectively.

Although different materials such as metals, ceramics and carbon can be used as a matrix for composites, polymers are the most widely used matrix materials, mainly because of their ease of fabrication. Polymers are long-chain molecules composed of many repeating units. They are organic in nature and commonly made of a carbon backbone, hydrogen and other non-metallic elements. When considering polymers two types can be distinguished: thermoplastics and thermosets [94,95,98]. Thermoplastics or thermosoftening plastics are polymers that soften during heating and harden upon cooling. This process can be repeated without limitation. Monofunctional monomers form a linear polymer network [94]. Their linear, non-crosslinked network gives them interesting properties. Although the elastic modulus of thermoplastic polymers is lower than that of thermosetting polymers, they are less brittle and offer higher toughness. An example of a dental thermoplastic is PMMA. Thermosets or thermosetting polymers cure in an irreversible way by heat application or through a chemical reaction [95,98]. Multifunctional monomers form a highly crosslinked three dimensional polymer network during the curing process [94]. An example of a dental thermoset is Bis-GMA also known as Bowen resin [72].

Different polymer matrices were proposed to be used in dental glass fibre-reinforced composites. Initial experiments screened for possible thermoplastic polymers such as polycarbonate [30,32,36], Poly(ethylene terephthalateglycol) or PETG [29,30,32,33,36], Poly(1,4-cyclohexylene dimetylene terephtalate glycol) PCTG [29], nylon-6 [30], nylon-12 [30,32], polyurethane [30] and polypropylene [36]. Further research focussed on the use of thermosetting polymers such as bisphenol-A-diglycidyl dimethacrylate/polyethylene glycol dimethacrylate or Bis-GMA/PEGDMA blends [34]. The major problem with thermoplastic and some of the dimethacrylate formulations were related to their bonding properties to particulate filler composites and to their handling properties [31]. Today, the most widely used polymers in dental glass fibre-reinforced composites are dimethacrylates and epoxies. Dimethacrylates are mainly used in uncured preimpregnated FRC used for fabrication of FDPs and periodontal splints, while epoxies are utilised in cured preimpregnated FRC such as root canal post. Well-known polymer formulations used for producing dental FRCs are bisphenol-A-diglycidyl dimethacrylate/triethylene glycol dimethacrylate or Bis-GMA/TEGDMA [99], bisphenol-A-diglycidyl dimethacrylate/urethane dimethacrylate or Bis-GMA/UDMA, urethane tetramethacrylate or UTMA [100] and copolyamide of PA66 and PA6T/6I [101]. A special group of polymer formulations are those forming an Interpenetrating Polymer Network (IPN). An IPN is a network formed by combining two or more polymers, which do not merge by chemical reaction but by interpenetration [102]. For dental FRCs only semi-IPNs are utilised, which means that one or more polymers are cross-linked and one or more polymers are linear [102]. In case of a dental semi-IPN the crosslinked part is formed by dimethacrylates, while the linear part is formed by the monofunctional methylmethacrylate [102]. A commercially available example of a semi-IPN-based FRC is everStick (Sticktech ltd, Turku, Finland) which contains a PMMA/Bis-GMA matrix. Semi-IPN matrices are used in favour of crosslinked thermoset matrices because they exhibit increased toughness, improved handling properties and superior mechanical interlocking of adhesives to IPN-like polymers [102].

Fibres

The filler fraction of a dental FRC consists mainly of fibres whose main function is reinforcement of the composite structure. This filler fraction is at least 50 times stronger and 10-150 times stiffer than the matrix [95]. This difference can be nicely illustrated by comparing E-glass and PMMA. The reported tensile strength and elastic modulus of E-glass is 3400 MPa and 73 GPa, while only 40 MPa and 2 GPa for PMMA [96]. This means that E-glass is approximately 85 times stronger and 36 times stiffer than PMMA.

The fibres used in dental FRCs can be of organic or inorganic nature. Both organic as well as inorganic fibres can be of natural or synthetic origin. Examples of organic synthetic fibres are ultra-high modulus polyethylene, aramide (Kevlar), and carbon. Glass and metals are representatives of the group of inorganic natural fibres. FRC are in more than 90% of the cases reinforced with glass fibres (Figure 1.8), which makes them the most widely used reinforcement. The main reason for choosing glass fibres in case of dental FRCs are the aesthetic properties.

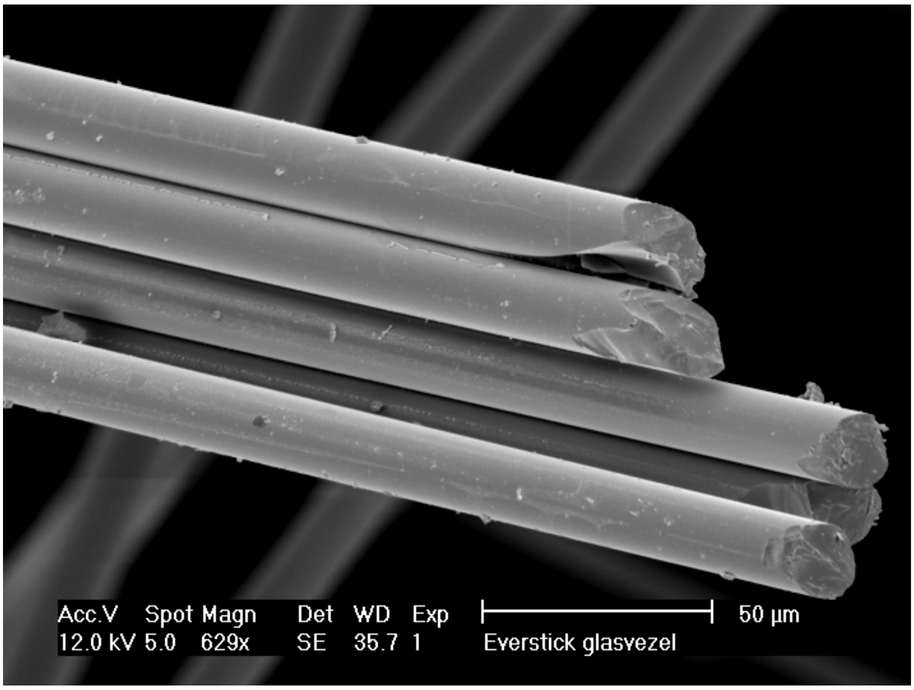

Figure 1.8 Scanning electron micrograph, 629x, of unidirectional E-glass fibres.

The basic component of glass is siliciumdioxide or silica (SiO_2). Silica is a polymorph material that can exist in many different forms. Silica can form either crystalline solids such as quartz, tridymite, cristobalite, or amorphous solids such as glasses. While quartz contains 99.95% SiO_2 in crystalline form [98], glasses are mixtures of SiO_2 and various oxides. Glasses can be tailored by adjusting their composition to meet specific requirements. Two categories of commercial glass fibres can be distinguished; low-cost general purpose fibres and premium special purpose fibres. Two generic types of general purpose glass fibres are available: boron-containing and boron-free E-glass. Several special-purpose fibres are of interest today, including ECR-glass fibres with high corrosion-resistance, S- or R-glass fibres exhibiting high strength, D-glass fibres with low dielectric constants, and pure silica or quartz fibres usable at ultrahigh temperatures. An overview on the composition and properties of the different glass fibre categories is presented in Table 1.1. Although quartz fibres or S-glass fibres are used for reinforcing dental FRCs, boron-containing E-glass fibres are still the most widely used glass fibre in dentistry.

Tabel 1.1 Composition and properties of commercial glass fibres.

Material	composition	Density (kg/m^3)	Tensile strength (MPa)	Elastic modulus (GPa)
boron-containing E-glass	52-56% SiO_2, 4-6% B_2O_3, 12-15% Al_2O_3, 21-23% CaO, 0.4-4% MgO, 0.2-0.5% TiO_2, 1% Na_2O, 0.2-0.4% Fe_2O_3, 0.2-0.7% F_2	2540-2550	3100-3800	76-78
boron-free E-glass	59% SiO_2, 12.1% Al_2O_3, 22.6% CaO, 3.4% MgO, 1.5% TiO_2, 0.9% Na_2O, 0.2% Fe_2O_3	2620	3100-3800	80-81
S-glass R-glass	60-65.5% SiO_2, 23-25% Al_2O_3,0-9%CaO, 6-11% MgO, 0-1% Zr_2O_3, 0-0.1% Na_2O, 0-0.1 Fe_2O_3	2480-2490	4380-4590	88-91
ECR-glass	58.2% SiO_2, 11.6% Al_2O_3, 21.7% CaO, 2.0% MgO, 2.9% ZnO, 2.5% TiO_2, 1.0% Na_2O, 0.2% K_2O, 0.1% Fe_2O_3	2660-2668	3100-3800	80-81
D-glass	74.5% SiO_2, 22% B2O3, 0.3% Al_2O_3,0.5%CaO, 2.9% ZnO, 2.5% TiO_2, 1% Na_2O, 0.2% K_2O, 0.1 Fe_2O_3	2160	2410	
Silica Quartz	99.99%SiO_2	2150	3400	69

1.5 Fabrication of dental fibre-reinforced composites.

The first step in the production of fibre-reinforced composites is blending the raw materials together. Blending of reinforcing fibres or fillers, resin matrix and additives can be done during different stages of the fabrication process [95]: before or during the shaping process. When the constituents are blended together before the shaping process, this is done during an additional stage called compound construction. During the stage of compound construction the constituents are mixed into a preliminary form that is suitable for shaping the FRC end-product. This preliminary form or shaping form contains the FRC in an uncured state and can be delivered as two types of forms: moulding compound or pre-impregnated compound, the so-called prepreg. Moulding compounds or prepregs are processed into the desired shape during the tooling stage of the final fabrication process. Most resin-based composites for dental use are delivered to the end-user as moulding compound or prepreg. Final shaping and curing is executed by the dentist or dental technician. On the other hand, if the constituents are combined during the shaping process then they are mixed into their final shape and leave the fabrication process cured into their final shape as an end-product. Only few dental FRCs such as fibre posts are delivered to the dentist or dental technician as an end-product.

Compound construction.

Moulding compounds [95,98].

Moulding compounds are fully-formulated materials that are primarily designed to be used during a moulding process. By definition, these materials get their final shape when placed into a mould by means of compression or injection. In order to comply with these needs they are uncured and should exhibit some degree of flow. Therefore the reinforcing fraction of moulding compounds are mainly short, randomly orientated fibres or particulate fillers. Dental resin-based composites such as particulate filler composites and short fibre composites are available for the practitioner as moulding compounds in pre-loaded tips and syringes. Moulding compounds can be categorised in three widely used industrial forms; (1) sheet moulding compounds, (2) bulk moulding compounds and, (3) thick moulding compounds. Only bulk moulding compounds will be briefly discussed here, since they are the only form used for dental purposes. Bulk moulding compounds consist usually of a thermosetting resin, fillers, short fibres and additives. The different components are mixed in a Z-blade mixer until everything is homogeneously distributed. Subsequently, the mixed composite or dough is extruded and loaded into syringes or pre-loaded tips (Figure 1.9).

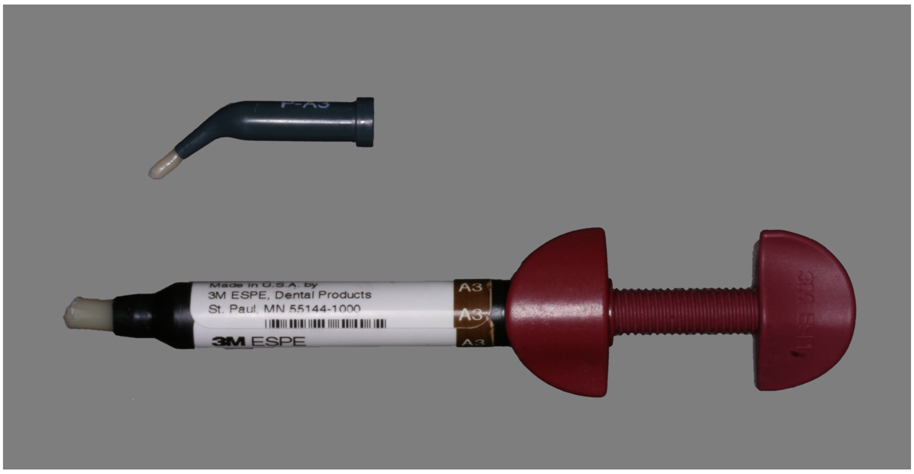

Figure 1.9 Packaging of commercially available particulate filler composites for dental use: pre-loaded tip (upper) and syringe (lower).

Prepregs [95,98].

Prepregs are resin pre-impregnated reinforcements. In contrast to moulding compounds, different forms of fibre reinforcement such as continuous unidirectional fibre bundles, woven mats and braids can be used. The fibre reinforcements are impregnated with a controlled quantity of uncured resin. Two production methods are well established for the fabrication of prepregs: the hot melt method and the solvent impregnation method [95,98]. The hot melt method implicates that the viscosity of the resin needs to be reduced during impregnation of the fibre reinforcement. This reduction in viscosity is only needed for a short time and is obtained with heat. Proper impregnation of the fibres with the heated resin is achieved under pressure. For the solvent impregnation method the viscosity of the resin is reduced by adding solvent. After the fibre reinforcement passed through the resin bath the solvent is evaporated in an oven. Dental FRCs using continuous or woven fibres are frequently supplied as prepregs (Figure 1.10). The popularity of prepregs can be attributed to the fact they exhibit superior fibre impregnation and their ease of use during shaping of the end-product.

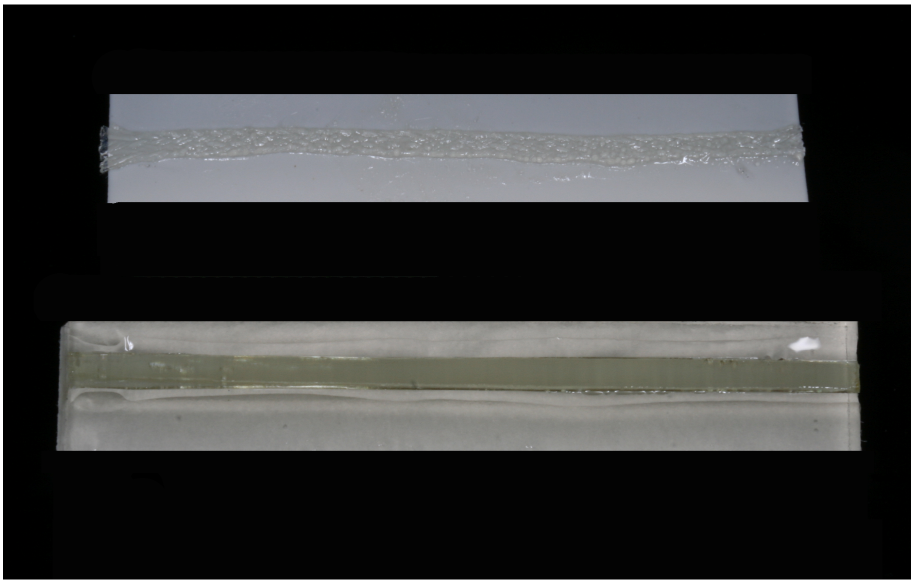

Figure 1.10 Commercially available pre-impregnated fibre-reinforced composites or prepregs for dental use: Interlig® (upper) and everStick C&B® (lower).

Fabrication process.

Final shaping and curing of FRC appliances is accomplished during the fabrication process. Industrial FRCs are in contrast to dental FRCs usually fabricated during an automated or semi-automated process in a factory [98]. The only dental

FRCs entirely made in an industrial process are fibre posts, which are fabricated by pultrusion. Pultrusion as a fabrication process is capable of producing fibre-reinforced composites with a constant cross-sectional shape in a continuous way [98]. It is the only fabrication process suitable for producing high-quality appliances in an almost completely automated process [95]. During the production process the fibre reinforcement is pulled through a resin impregnation system. Subsequently the resin impregnated fibres are passing through preforming guides in order to assemble the fibres and remove the excess resin. Afterwards the pre-shaped FRC is pulled through a heated die where final shaping, compaction and curing are performed. Pultrusion makes it also possible to pre-stress the fibres in order to obtain FRCs with improved mechanical properties [103].

On the other hand, dental FRC restorations are manually fabricated direct into the mouth by the dentist or in the dental laboratory by a dental technician. Most FRC frameworks are fabricated by hand lay-up technique and polymerised with light, light and heat, or light and vacuum. Only two commercially available dental FRC-systems make use of a closed mould process to fabricate their FRC frameworks. In case of the Vectris system (Ivoclar-Vivadent, Schaan, Liechtenstein) the FRC prepregs are manually packed into a translucent silicon mould. This silicon mould is positioned over the gypsum working model and placed in a fully automated framework former (Vectris VS1; Ivoclar-Vivadent), where the FRC framework is formed under pressure and subsequently polymerised with light.

1.6 Principles of fibre reinforcement.

The effect of fibre reinforcement is depending on several factors which strongly affects the mechanical properties of a fibre-reinforced composite:

1) Properties of fibres and matrix
2) Quantity of fibres
3) Orientation of fibres
4) Impregnation
5) Adhesion
6) Fibre diameter

Properties of fibres and matrix.

The mechanical properties of a fibre-reinforced composite are primarily determined by its components: the materials used for the fabrication of the fibres and the matrix. The mechanical properties of a composite tend to be intermediate between those of the components and follow quiet often the law of mixtures [104]. The theoretical strength (tensile strength) and stiffness (elastic modulus) of a composite can be calculated according to the volume fraction of the fibres and the matrix. Although one should be aware of the fact that the presented formulas only apply when the load is applied parallel to the fibre direction. Theoretical elastic modulus of a composite (*E_c)* according to the law of mixtures [96] can be calculated with the following formula:

$$E_c = E_f V_f + E_m V_m$$

where E_f is the elastic modulus of the fibres, V_f is the volume fraction of the fibres, E_m is the elastic modulus of the matrix and V_m is the volume fraction of the matrix. The theoretical tensile strength of a composite (σ_c) according to the law of mixtures [96] can be calculated with the following formula:

$$\sigma_c = \alpha V_f \sigma_f + V_m \sigma_m$$

where α is the reinforcing efficiency factor or Krenchel factor, V_f is the volume fraction of the fibres, σ_f is the tensile strength of the fibres, V_m is the volume fraction of the matrix and *m* is the tensile strength of the matrix.

Other properties, like toughness, do not follow the law of mixtures and can be significantly superior to those of both individual components.

Quantity of fibres.

The fibre quantity of a FRC can be expressed in two ways: in fibre weight fraction (weight %) and in fibre volume fraction (volume %). The most straightforward and comparative way to present the fibre quantity of an FRC is as fibre volume fraction, since different fibres exhibit different densities. Fibre volume fraction can be calculated with the following formula:

$$V_f = (W_f / \rho_f) / (W_f / \rho_f + (1 - W_f) / \rho_r)$$

where W_f is the weight proportion of the fibres, ρ_f the density of the fibres and ρ_r the density of the resin matrix.

Several studies demonstrated the effect of fibre quantity on the mechanical properties of FRC. Increase in fibre volume fraction results in higher tensile strength

[96], flexure strength [68,105,106], elastic modulus [29,107], impact strength [108] and toughness [107].

Fibre orientation.

The mechanical properties of fibre-reinforced composites are strongly influenced by the orientation of the fibres in relation to the direction of the load. The effect of fibre orientation was first described by Krenchel and can be presented as reinforcing efficiency factor or Krenchel's factor [109]. The most efficient reinforcement can be obtained with continuous unidirectional fibres, which give anisotropic properties to the FRC [110]. This means that optimal reinforcement (Figure 1.5) can only be obtained when the direction of the stress is parallel to the direction of the fibres, meaning that the fibres are loaded in tension [111]. Bi-directional fibre weaves reinforce a material in two directions and give orthotropic properties to the FRC [110,111]. Although the FRC is reinforced in different directions, the reinforcing efficiency is still dependent on the direction of the applied stress to the direction of the fibres (Figure 1.5). FRCs with short fibres in random directions exhibit isotropic properties, meaning that the fibre reinforcement is equal in all directions (Figure 1.11) [110,111].

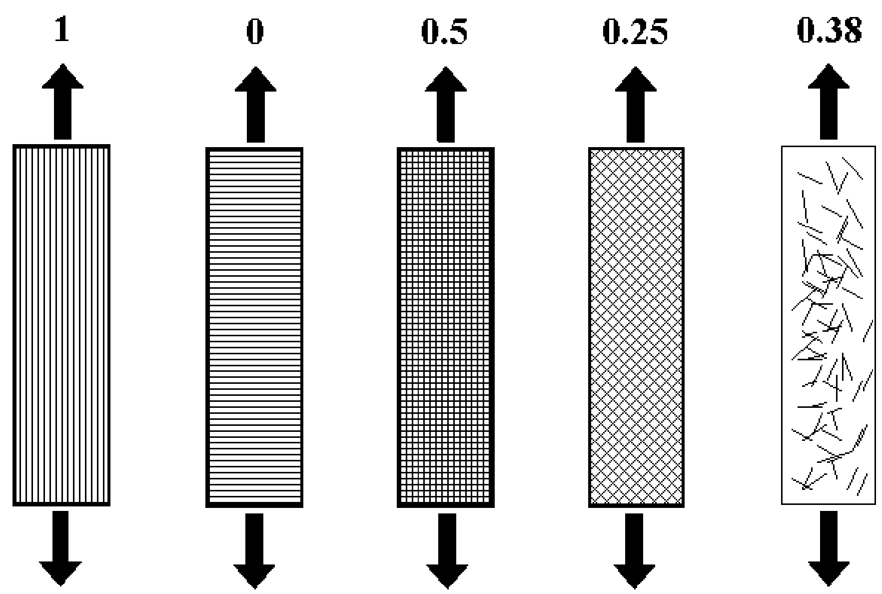

Figure 1.11 Reinforcing efficiency (number above drawings) of fibres depending their orientation in relation to the direction of the load. The load direction is shown by the black arrows.

Impregnation.

Fibre-reinforced composites only have ideal mechanical properties if stresses are optimally transferred from resin matrix towards fibres. Fibres should be thoroughly impregnated with the resin to assure in the first instance adhesion and in the second instance stress transfer. Incomplete impregnation of the fibres prevents transfer of shear stresses from the matrix to the fibres [112] and therefore can explain the negative discrepancy found between theoretical properties and the experimental properties [96]. Two problems are associated with void formation within incompletely impregnated FRC: increased watersorption [113] and inhibition of polymerisation [114]. Both problems reduce the mechanical properties of the FRC [113,115]. Several studies pointed out that complete wetting of the fibres is difficult to achieve with manual impregnation techniques [96,106,116]. Therefore, pre-impregnated FRCs were brought to the attention to eliminate the problem of incomplete impregnation. Goldberg *et al*. [29,30,117] introduced, already in the early 1990s, a pre-impregnated FRC made of S-glass fibres and Bis-GMA resin matrix. In order to obtain proper impregnation of the glass fibres, the FRC was fabricated by pultrusion [29]. Vallittu *et al*. [118,119] developed, in the late 1990s a FRC-system in which the glass fibres were pre-impregnated with highly porous PMMA. This FRC-system needed further manual impregnation during the fabrication of the dental restoration. In the beginning of the new millennium a fully pre-impregnated FRC, with improved handling properties, was introduced [111] consisting of E-glass fibres impregnated with a polymer-monomer gel consisting of Bis-GMA and PMMA

Adhesion.

Once again adhesion between fibres and matrix is of paramount importance in order to ensure efficient matrix-fibre stress transfer. Proper adhesion between glass and resin is not evident, but desirable to achieve sufficient strength, since both materials are chemically inert to each other [59]. Reliable chemical bonding between glass fibres and resin matrix by forming covalent bond can be achieved by the use of silane coupling agents [120,121]. The most commonly applied silane in dentistry is the monofunctional γ-methacryloxypropyltrimethoxysilane or γ-MPS [122]. Silanes are bifunctional molecules, meaning that they exhibit dual reactivity. The organic functional end can copolymerise with the organic resin matrix by forming an ester linkage [122,123]. The alkoxy group on the other functional end can react with hydroxy groups on glass and silica fibres to form a polysiloxane network (Si-O-Si) [122,123]. Notwithstanding the good silane-promoted adhesion obtained between glass fibres and polymer matrix, one should be aware that the polysiloxane network on the glass fibres surface is prone to hydrolysis by water [123-125].

Fibre diameter.

Little is known about the effect of fibre diameter on the mechanical properties of fibre-reinforced composite. A recent study by Obukuro *et al*. [126] revealed that flexural properties of an E-glass/urethane dimethacrylate-triethylene glycol dimethacrylate composite were affected by the fibre diameter. This study showed that flexural strength was affected by the fibre diameter, but no effect was observed regarding elastic modulus. Highest flexural strength was obtained for fibre-reinforced composite with continuous unidirectional fibres with a diameter ranging from 20-30 μm [126].

1.7 Commercially available dental glass fibre-reinforced composites.

Today, several glass fibre-based dental FRC-systems are commercially available. An overview of the most widely used pre-impregnated FRCs is given in Table 1.2.

Tabel 1.2 Classification of pre-impregnated glass fibre-reinforced composites.

Brand	Composition Fibre	 Matrix	Fibre orientation	Manufacturer
everStick	E-glass	Bis-GMA, PMMA	unidirectional	Sticktech Ltd, Turku, Finland
Stick	E-glass	PMMA	unidirectional	Sticktech Ltd, Turku, Finland
Fibrekor Splint-it	S-glass	Bis-GMA	unidirectional	Pentron, Wallingford, CT, USA
Vectris	R-glass	Bis-GMA, TEGDMA	unidirectional woven	Ivoclar-vivadent, Schaan, Liechtenstein
EGfiber	E-glass	UTMA	unidirectional	Kuraray medical inc, Osaka, Japan.
Fibrex-lab	S-glass	Bis-GMA, UDMA	unidirectional multidirectional braided	Angelus, Londrina, PR, Brazil
Interlig	S-glass	Bis-GMA, UDMA	intertwined	Angelus, Londrina, PR, Brazil
Tender fiber	Glass	Bis-GMA	unidirectional	Micerium, Avegno, Italy
Quartz splint	Quartz	Bis-GMA	unidirectional woven and mesh rope	RTD, St-Egrève, France

1.8 Properties of fibre-reinforced composites

Flexural strength and modulus.

Recently, the literature on *in vitro* tests of FRC beams was reviewed and analysed by means of meta-regression [127]. Included studies were performed according to the ISO 4049 protocol and compared longitudinal reinforced specimens to a control group of unreinforced specimens. All included studies reported on the flexural properties of resin-based composite beams reinforced with fibres placed in different locations. This systematic review confirmed the reinforcing effect of fibres on resin composite beams. The flexural strength of resin-based composite bars reinforced with fibres placed at the tensile side of the specimens varied between 185 MPa and 577 MPa. The incorporation of fibres increased the average flexural strength by 100 to 200 MPa. The flexural modulus of resin-based composite bars reinforced with fibres placed at the tensile side varied between 2 GPa and 15 GPa. The data showed that increase as well as decrease of flexural modulus is noted by incorporation of fibres. One should keep in mind that the reported average values are calculated of pooled data from glass-reinforced, as well as polyethylene-reinforced specimens. It is well known that the reinforcing effect of polyethylene fibres is inferior to that of glass fibres [59,128-130]. Ellakwa *et al*. [128] found a significant increase in flexural strength for glass fibres in comparison to polyethylene fibres, 102 MPa and 301 MPa, respectively. A study by Lassilla *et al*. [131] noted a difference in flexural strength of 454 MPa between unreinforced and glass fibre-reinforced beams. It is generally accepted that the incorporation of glass fibres increases the flexural modulus of composite beams [97,107,128,132]. One the other hand, incorporation of polyethylene fibres does not significantly affect the flexural modulus of composite beams [67] or even has a tendency to decrease the flexural modulus [97,130,133]. The difference between polyethylene and glass fibre is nicely illustrated by the study of Ellakwa *et al*. [128]. A significant increase in flexural modulus up to 5.1 GPa was found for glass fibres, while polyethylene fibres only increased the flexural modulus by 1.8 GPa if wetted by a filled bonding agent.

The flexural strength and modulus of plain glass fibre-reinforced composite outperforms the values of combination beams discussed in the previous paragraph. Several studies investigated some of the commercially available FRC, such as Fibrekor, Vectris, everStick and EGfiber. The literature reports flexural strength values for Fibrekor, everStick and EGfiber that vary from 367 MPa to 1201 MPa [99,132,134,135], 559 MPa to 1164 MPa [100,136] and 547 MPa to 689 MPa

[100,132]. On the other hand the values for Vectris lies closer together and only vary between 618 MPa and 696 MPa [99,132,135]. Flexural modulus values ranging between 19 GPa and 30.2 GPa are reported for Vectris [99,132], 22.3 GPa and 26.7 GPa for FibreKor [99,132], 23.8 GPa and 26.7 GPa for everStick [100,125] and, 24.2 GPa and 25.5 GPa for EGfiber [100,132].

Fatigue resistance.

Fatigue resistance is probably the most important and clinically relevant property of a dental material. Especially, since dental reconstructions tend to fail in the majority of the cases because of mechanical fatigue [137]. During fatigue loading materials are repeatedly subjected to a stress below the yield stress of the material, which eventually reduces the strength of the material and will cause failure. Therefore, fatigue can be defined as a progressive fracture under repeated loading [138]. The fatigue resistance or strength of a material is the stress at which failure occurs under repeated loading and is dependent on the magnitude of the load and the number of load applications [138].

Notwithstanding the importance attributed to fatigue, little information about the fatigue resistance of dental FRCs is available [139-144]. Bae *et al*. [143] studied the dynamic fatigue strength of bar-shaped specimens made of a combination of FRC and PFC and tested in a three-point bending mode according to the staircase method. The fatigue strength at 10^5 cycles was determined. They concluded that fibre reinforcement had a beneficial effect on the fatigue strength of resin composites. The fatigue strength of unreinforced specimens varied from 49 MPa to 57 MPa, while the fatigue strength of fibre-reinforced specimens varied from 90.2 MPa and 196.9 MPa. Narva *et al*. [144] studied the flexural fatigue behaviour of fibre-reinforced PMMA-based denture base resin. Bar-shaped specimens were subjected to a constant-deflection fatigue test in a cantilever beam test set-up for a maximum of 10^5 cycles. They observed a significant increase in mean number of loading cycles for fibre-reinforced specimens in comparison to unreinforced specimen. A second study by Narva *et al*. [140] reported on the fatigue resistance of cylindrical test specimens entirely made of FRC. The fatigue resistance was determined by a constant-deflection fatigue test for a maximum of 150,000 loading cycles. Non of the specimens fractured after 150,000 cycles. However, the mean force to cause a deflection of 1 mm was significantly reduced from 33.5 N towards 23.4 N for dry-stored specimens and from 37.7 N down to 13.1 N for water-stored specimens. Drummond *et al*. [139] evaluated the flexure strength of bar-specimens made of FRC under static and cyclic loading. A reduction in strength of up to 38% was observed for specimens subjected to 1,000 loading cycles.

Baran *et al*. [142] stated in their review that the fatigue resistance of resin composite reinforced with glass fibres was increased because fibre reinforcements dissipates the stresses generated by the applied loads and they are able to arrest and/or deflect cracks. It was brought to the attention that there exists a difference in fatigue related damage between FRCs with a high fibre volume fraction and those with a low volume fraction [145]. In FRCs wit a high fibre content, fatigue behaviour is dominated by the fibre properties when high stresses are applied, while matrix-related damage occurs when low stress are applied. In case of FRCs with low fibre content the fatigue related damage is strictly matrix-related.

1.9 Aims of the thesis

Although the science of fibre-reinforcement is well established in the field of engineering, the application of fibre-reinforced composites in dentistry is relatively new. Several aspects regarding material properties, framework design and indication are still not well understood.

The aim of this thesis was to explore *in vitro* the influence of fibre-reinforcement on certain mechanical properties and prosthesis designs.

The specific aims were:

1. To evaluate *in vitro* the influence of fibre-reinforcement on the fracture strength and fatigue resistance of resin-based composites.
2. To evaluate *in vitro* the influence of fibre-reinforcement and luting cement on the static failure load (SFL) and dynamic failure load (DFL) of two-unit cantilever RB-FDPs by using simplified cantilever beams.
3. To compare, by means of three-dimensional Finite Element Analysis, the biomechanical behaviour of anterior two-unit cantilever RB-FDPs made of various framework materials.
4. To investigate *in vitro* the influence of retainer design on the strength and stress distribution in the tooth/restoration complex of indirect two-unit cantilever RB-FDPs in the premolar region.
5. To evaluate *in vitro* the influence of framework design on the load-bearing capacity of laboratory-made inlay-retained FRC-FDPs.

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CHAPTER 2

Fracture strength and fatigue resistance of dental resin-based composites

2.1 Abstract

Objectives: The aim of this study was to evaluate *in vitro* the influence of fibre-reinforcement on the fracture strength and fatigue resistance of resin-based composites.

Materials and Methods: One hundred rectangular bar-shaped specimens (2 x 2 x 25 mm) made of resin-based composite were prepared in a stainless steel split-mould: (i) thirty specimens of particulate filler composite (PFC) (Filtek Z100, 3M ESPE, St Paul, MN, USA), (ii) thirty specimens of fibre-reinforced composite (FRC) (everStick C&B, Sticktech Ltd, Turku, Finland) and (iii) forty specimens of PFC and FRC combined in two longitudinal layers of equal thickness. Each specimen was trimmed into a cylindrical hourglass shape. The fracture strength (cantilever beam test, n = 10) and the fatigue resistance (rotating cantilever beam test; staircase method: 10^4 cycles, 1.2 Hz, n = 20) were determined. Fracture strength, fatigue resistance and work-of-fracture were calculated. The fracture surfaces of failed specimens were analysed with SEM. Data was analyzed by logistic regression, one-way ANOVA followed by Tukey's post hoc test and, a student t-test.

Results: ANOVA revealed that fibre-reinforcement had significant effect ($p < 0.001$) on fracture strength, fatigue resistance, and work-of-fracture. Student t-test showed significant differences ($p <0.001$) in fatigue resistance compared to fracture strength.

Conclusions: Within the limitations of this study, the following conclusions can be drawn (i) the fatigue resistance of resin-based composites is lower than their fracture strength and (ii) FRC are more fatigue resistant than PFC or combinations of FRC and PFC.

2.2 Introduction

The use of resin-based composites increased enormously during the last two decades. Their increasing popularity could be attributed to the paradigm shift from G.V. Black's "extension for prevention" [1] to minimal invasive dentistry [2] established by the development of adhesive dentistry. The adhesive revolution and subsequent popularity of resin-based composites was initiated by two major breakthroughs: the introduction of the acid etch technique by Buonocore in the mid 1950s [3] and the development of Bis-GMA as an organic matrix for resin composites by Bowen in the early 1960s [4]. When considering resin-based composites one should keep in mind that this in fact represents a composite family consisting of, among others, particulate filler composites and fibre-reinforced composites, being the subject of the present study.

The range of indications where resin-based composites in general and particulate filler composites (PFC) in particular are used has expanded explosively due to their enhanced physical and mechanical properties. Today, resin-based composites are indicated on a regular basis for posterior direct and laboratory made restorations, as an extension to their original indication which was limited to direct restorations in anterior teeth. After the introduction of fibre-reinforced composites (FRC) and especially the development of glass fibre-reinforced composites [5], resin-based composites came into focus as a material that has the capabilities to be used for the fabrication of fixed dental prosthesis (FDP). In order to be able to withstand the chewing forces, resin-based composite FDPs are made of a FRC-framework veneered with PFC, with FRC acting as a stress dissipater while the PFC gives the construction its aesthetic properties. This type of prosthetic constructions is known in literature as fibre-reinforced composite fixed dental prostheses (FRC-FDPs). The growing interest in this type of restorations was stimulated by the high demand for improved aesthetics and by the growing concerns related to metallic restorations [6].

In spite of the less favourable longevity exhibited by cantilever FDPs in comparison to fixed-fixed FDPs [7,8], there is still a persistent need for this treatment option. The mostly used indication for cantilever FDPs is for extending a shortened dental arch. Also in the field of resin-bonded FDPs a two-unit cantilever design can be a viable alternative that has proven to perform as well as or even better than their three-unit fixed-fixed counterparts [9-11]. The clinical success of two-unit cantilever resin-bonded FDPs led to the use of resin-based composites for the fabrication of these restorations. Two-unit cantilever resin-bonded FRC-FDPs are already used for single tooth replacement in the anterior region [12-14] and maybe in the future also in the posterior region [15].

Dental restorations during clinical functioning are not only subjected to high static loads, but also to low cyclic loads, the latter known as fatigue loading. Fatigue is a mode of failure whereby cracks are induced by subjecting a material or structure to repeated sub-critical loads, which leads eventually to failure [16]. Mechanical failure of dental restorations can be attributed in the majority of the cases to fatigue loading, which makes fatigue resistance one of the most important and clinically relevant properties of a dental material or restoration [16]. Little information about the fatigue resistance of fibre-reinforced composites used in dentistry is available at the moment [17-20]. The forces occurring during physiological function have a vertical as well a horizontal component, which make them multi-vectorial in nature [21]. A representative laboratory test should cope with both aspects, which can be accomplished by means of a rotating cantilever beam fatigue test [16].

The aim of this study was to evaluate *in vitro* the influence of fibre-reinforcement on the fracture strength and fatigue resistance of resin-based composites. The fracture strength was obtained by means of a cantilever beam test, while the fatigue resistance was obtained according to the staircase approach in a rotational cantilever beam fatigue testing device. The null hypothesis to be tested in this experiment was that a fibre-reinforced composite exhibited a comparable fatigue resistance as a particulate filler composite.

2.3 Materials and Methods

Table 2.1 Materials used in the study.

Brand	Composition	Manufacturer	Lot number
Filtek Z100	Resin: Bis-GMA, TEGDMA; Filler: zirconia, silica (≈ 64.2 vol%)	3M-ESPE Dental products, St Paul, MN, USA	70-2010-2226-9
everStick C&B	Resin: PMMA, Bis-GMA; Filler: silanised E-glass fibres (≈ 65 vol%)	Sticktech Ltd., Turku, Finland	2070212-ES-179

Bis-GMA bisphenol-A-glycidyl dimethacrylate; TEGDMA triethylenglycol dimethacrylate; PMMA poly(methyl methacrylate).

Two resin-based composites, *i.e.* one particulate filler composite (PFC) (Filtek Z100, 3M ESPE, St Paul, MN, USA) and one fibre-reinforced composite (FRC) (everStick C&B, Sticktech Ltd, Turku, Finland), both within their field of application widely used and Bis-GMA-based materials, were selected for this experiment.

EverStick C&B was delivered as prepregs containing 4000 continuous unidirectional silanised E-glass fibres (≈ 65 vol%) of 21 µm in diameter impregnated with light polymerisable semi-interpenetrating polymer network of PMMA/Bis-GMA resin. The composition of the materials used is summarised in Table 2.1.

Specimen preparation

One hundred rectangular bar-shaped specimens (2 x 2 x 25 mm) made of resin-based composite were prepared in a stainless steel split-mould. Thirty specimens were made of PFC, thirty specimens were made of FRC and forty specimens were made of a combination of FRC and PFC in two longitudinal layers of equal thickness.

To prepare the specimens the mould was filled in bulk with a resin-based composite (PFC, FRC or a bilayer of FRC and PFC) and covered on both sides with a cellophane sheet and a slide glass. In order to fabricate bilayered specimens with layers of equal thickness, the mould was first filled with a 1 mm thick layer of FRC and subsequently the mould was completely filled by adding PFC. The thickness of the FRC layer was checked with a Teflon space maintainer. The specimens were light cured for 60 s (3 x 20 s overlapping irradiation) on each side by a handheld polymerization unit (Astralis 10, Ivoclar-Vivadent, Schaan, Liechtenstein) with a power output of 1000 $mW \cdot cm^{-2}$ (Curing Radiometer model 100, Demetron Research Corporation, Danbury, USA).

The specimens were mounted in a lathe cutting machine (Micro miller MF 70, Proxxon GmbH, Niersbach, Germany) and trimmed under continuous water-cooling at one-third of their length into a cylindrical hourglass shape with a diameter of 1.2 ± 0.1mm using a tungsten carbide bur (H79EF.104.040, Komet, Lemgo, Germany). The diameter of the hourglass-shaped constriction was measured by a laser micrometer (Laser Scan Micrometer LSM 6000/LSM 503, Mitutoyo, Kawasaki, Japan). All specimens were stored in 37°C distilled water for at least 72 h until testing.

Fracture strength

The fracture strength was obtained by subjecting the specimens (n = 10) to a cantilever beam test (Figure 2.1A). Bar specimens were fixed in a custom-made device in a way the clamps bordered the hourglass shape. Four groups of 10 specimens each were tested:

1. PFC: specimens made of PFC only.
2. FRC: specimens made of FRC only.
3. FRC-t: bilayer specimens made of a combination of PFC and FRC, where the FRC is placed at the tension side of the specimens.

4. FRC-c: bilayer specimens made of a combination of PFC and FRC, where the FRC is placed at the compression side of the specimens.

The load was applied at a distance of 10 mm from the hourglass-shaped constriction by a steel rod. The specimens were loaded till failure in a universal testing machine (Hounsfield 12B AD, model 20-30, Salfords, UK) at a cross-head speed of 0.5 mm·min^{-1} and data were recorded by PC software (ACTA inTense 3.15, ACTA, Amsterdam, the Netherlands).

The fracture strength (S, MPa) of the first two groups, which were made of only one material, was calculated with the formula found in most textbooks on engineering science [22]:

$$S = \frac{32Fl}{\pi d^3} \tag{1}$$

The load F (N) multiplied by the distance l (10 mm) between the point of loading and the hourglass constriction represent the applied moment, and d (mm) is the smallest diameter of the constriction.

Because of the different elastic modules of the bilayer groups (FRC-t and FRC-c) the neutral line [22] is not at the centre of the cross-section and therefore equation 1 cannot be used. The shift of the neutral line (n, mm) relative to the centre of the cross-section is calculated as

$$n = \frac{2d}{3\pi} \cdot \frac{1-R}{1+R} \tag{2}$$

$R = E_c$ / E_t, is the ratio between the elastic moduli of the materials of the compressive and tensile layers respectively. Because assessment of the elastic modulus more or less requires complex finite element analysis modelling and as only the ratio between these is used, the ratio (R) may be found with

$$R = \frac{E_c}{E_t} = \frac{F_c \cdot l_c^2}{D_c \cdot d_c^4} \cdot \frac{D_t \cdot d_t^4}{F_t \cdot l_t^2} \tag{3}$$

This equation uses the stiffness of one-material rods (F/D) to compare the moduli with a correction for the position (l) and diameter (d) of the constriction, where most of the bending occurs. With these definitions, the strength on the tensile side becomes

$$S = 192 \cdot F \cdot L \frac{d - 2n}{3\pi d^4 - 64nd^3 + 48n^2d^2 + R \cdot (3\pi d^4 + 64nd^3 + 48n^2d^2)} \qquad (4)$$

With equal modules E_t and E_c, R becomes 1, equation 2 returns n = 0 and equation 3 simplifies to equation 1. If the maximum compression on the opposite side is of interest, equation 3 is multiplied by -R and the minus sign in the numerator is replaced with a plus ($d+2n$).

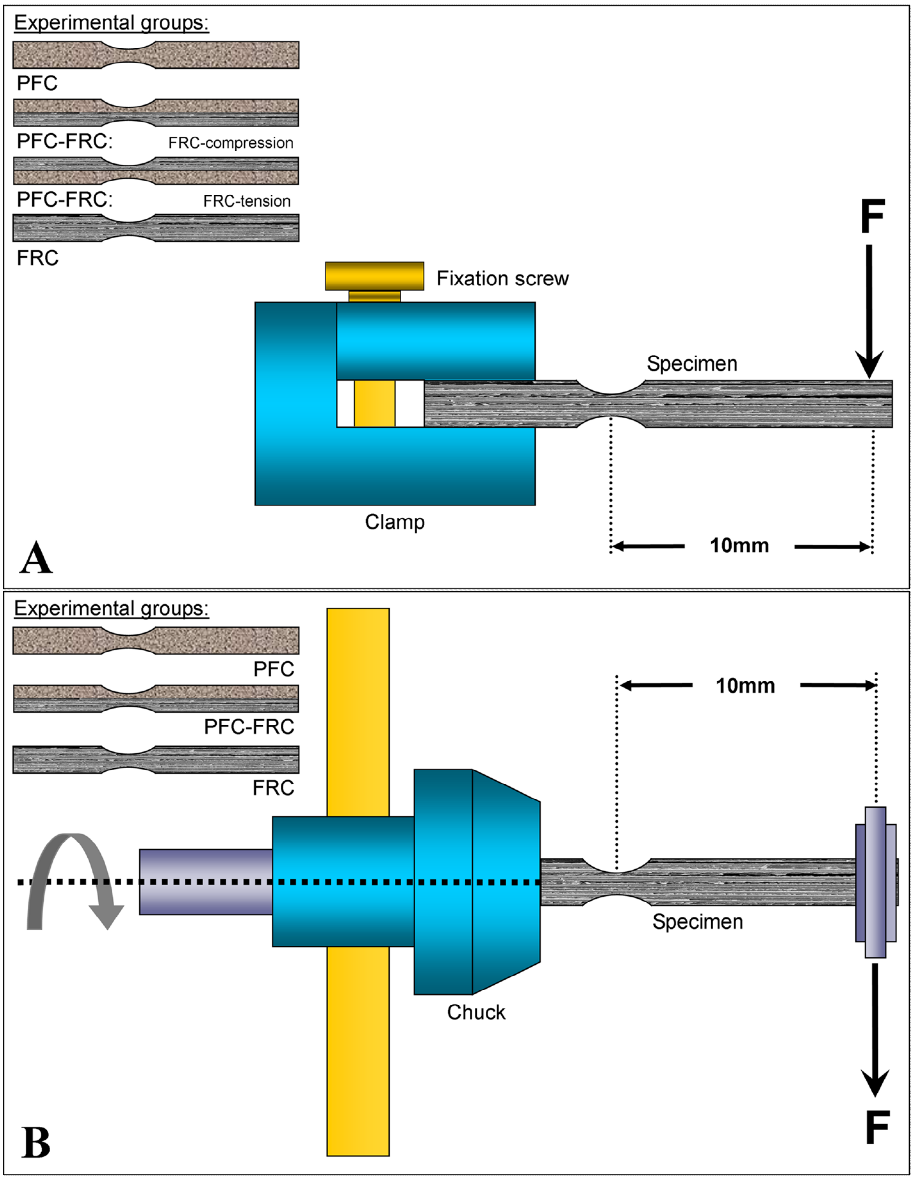

Figure 2.1 Schematic representation of (A) the cantilever beam test and (B) the rotating cantilever beam fatigue testing device: specimens are rotating around their longitudinal axis and loaded by attaching a weight (F) to a ball-bearing.

Work-of-fracture

The load-displacement curves were used to calculate work-of-fracture (γ_{WoF}, $kJ \cdot m^{-2}$) is the energy required to fracture a specimen and is calculated by dividing the area under the load-displacement curve by the specimen's cross-sectional area:

$$\gamma_{WoF} = \frac{4A}{\pi d^2} \qquad (5)$$

Where A (N·mm) is the area under the load-displacement curve and $\frac{1}{4}\pi\ d^2$ is the area of the cross-section (mm^2).

Fatigue resistance

The fatigue resistance of resin-based composites was determined by using a rotating cantilever beam fatigue testing device (Figure 2.1B). The specimens (n = 20) for each group; PFC, FRC, and FRC-PFC, respectively) were mounted in the chuck of the fatigue device. A ball-bearing was fixed to the beams at a distance of 10 mm from the hourglass constriction. The stress at the smallest diameter of the constriction was induced by attaching a weight onto the ball-bearing. The required force (F) was calculated with equation 1.

The "staircase" or "up-and-down" method was used as the analytical method for this experiment. Within each group of specimens the initial stress was set at *ca*. 50% of the previously determined fracture strength for all groups. The specimens were rotated at 1.2 Hz for 10^4 cycles or until failure. If failure occurred before 10^4 cycles the stress was decreased with 7.5% of the original fracture strength, respectively increased with the same percentage when the specimen survived.

Fibre volume fraction

The fibre volume fraction of each test group was determined by the resin burn-off method. The weight of the specimens (n = 3) was measured (Mettler AT261; Mettler Instrument, Highstone, NJ, USA) before and after combustion of the resin matrix for 1 h at 700ºC. The particulate fillers were mechanically removed. The fibre volume fraction V_f in vol % was calculated with the following formula:

$$V_f = \left(W_f / \rho_f\right) / \left(W_f / \rho_f + \left(1 - W_f\right) / \rho_r\right) \qquad (6)$$

where W_f is the weight proportion of the fibres, ρ_f the density of the fibres (E-glass = 2.54 $g \cdot cm^3$) and ρ_r the density of the resin matrix (everStick resin = 1.22 $g \cdot cm^3$).

Failure mode

All fractured specimens were visually examined and inspected under a light microscope (4 x magnification). A number of representative specimens (cantilever beam test: n = 3; rotating cantilever fatigue test: n = 6) were gold sputtered (Edwards Sputter Coater S150B, Edwards High Vacuum, Crawley, West Sussex, England) and their fracture surface was examined by scanning electron microscopy (Philips XL 20, Eindhoven, the Netherlands).

Statistical Analysis

Statistical analysis was performed with the statistical software SigmaStat 3.0 (SPSS Inc. Chicago, IL, USA). The dataset obtained from the fatigue test was analyzed using logistic regression in order to determine the fatigue resistance and standard deviation. The fatigue resistance can be defined as the load at which the probability of failure is 50%. Means and standard deviations of fracture strength, work-of-fracture and fatigue resistance for each group were calculated (Table 2.2). One-way analysis of variance (ANOVA) followed by Tukey's *post hoc* test were performed to determine the effect of fibre-reinforcement on the observed fracture strength, work-of-fracture and fatigue resistance. P-values of less than 0.05 were considered to be statistically significant. A student t-test was performed to compare the effect of fatigue testing and fracture testing.

2.4 Results

The mean fracture strength, fatigue resistance, work-of-fracture and fibre volume fraction of each experimental group are summarised in Table 2.2.

One-way ANOVA revealed significant differences in fracture strength ($F = 111.9$; $p < 0.001$), work-of-fracture ($F = 33.1$; $p < 0.001$), and fatigue resistance ($F = 436.7$; $p < 0.001$) of hourglass-shaped specimens made of different resin-based composites.

The highest fracture strength, work-of-fracture and fatigue resistance were obtained for FRC bars. Specimens made of a combination of FRC and PFC showed the second highest fracture strength and work-of-fracture when the FRC was placed in tension (FRC-t). There was no significant difference regarding fracture strength and work-of fracture between PFC bars and FRC-PFC combination bars when the FRC was placed in compression (FRC-c).

Student t-test showed statistically significant differences in fatigue resistance compared to fracture strength of hourglass-shaped specimens made of PFC ($t = 9.3$; $p < 0.001$), FRC ($t = 14.7$; $p < 0.001$) and bilayered specimen with FRC in tension ($t = 50.7$; $p < 0.001$).

Table 2.2 Mean fibre volume fraction, fracture strength, fatigue resistance, work-of-fracture and, ratio between fracture strength and fatigue resistance, with SD in parentheses, of resin-based composites. Test groups with the same superscript letter are not statistically different.

<table>
<tr><th>Group</th><th>Fibre volume fraction (vol %)</th><th>Fracture strength (MPa)</th><th>Fatigue resistance (MPa)</th><th>Work-of-Fracture ($kJ \cdot m^{-2}$)</th><th>Ratio</th></tr>
<tr><td>PFC</td><td>0</td><td>164.9^{a} (29.7)</td><td>51.5^{d} (32.3)</td><td>0.55^{B} (0.19)</td><td>0.32</td></tr>
<tr><td>FRC-t</td><td rowspan="2">16</td><td>539.5^{b} (35.2)</td><td rowspan="2">116.5^{a} (9.9)</td><td>14.20^{C} (2.21)</td><td>0.15</td></tr>
<tr><td>FRC-c</td><td>125.2^{a} (38.0)</td><td>0.21^{B} (0.13)</td><td>0.95</td></tr>
<tr><td>FRC</td><td>42</td><td>936.1^{c} (218.5)</td><td>231.9^{e} (2.9)</td><td>27.90^{D}(14.17)</td><td>0.25</td></tr>
</table>

Failure mode

SEM examination of the broken specimens demonstrated that PFC bars failed from a critical crack (point of highest stress concentration) that was located at the periphery of the beams (Figure 2.3A).These specimens revealed an elastic behaviour and were associated with an immediate drop in load once the ultimate strength was reached, defined as instantaneous failure (Figure 2.2C) [23]. The highest fracture strength was observed for the FRC specimens which exhibited a statistical failure demonstrated by a region of plastic deformation which was related to damage accumulation in the form of rupture of the fibres located in the tensile side of the specimens (Figure 2.3B), followed by a region of unstable condition [23]. This behaviour was observed as deviation from linearity observed on the load-displacement diagram (Figure 2.2A). For the bilayer specimen, two patterns of different behaviour were observed; (i) when the fibres where on the compressive side (FRC-c), the specimens failed at a low load comparable to that of specimens made of PFC alone. Additionally, there was a sudden drop in the applied load indicating fracture of the PFC-part, followed by a more or less horizontal component in the load-displacement curve where the remaining fibres carried the applied load (Figure 2.2D and Figure 2.3C). On the other hand, (ii) when the fibres were on the tensile side (FRC-t) the specimens failed at a significantly higher load and demonstrated behaviour more comparable to FRC specimens (Table 2.2 and Figure 2.2B). Nevertheless, all specimens demonstrated also compressive fracture of the PFC (Figure 2.3D).

SEM analysis of FRC-PFC specimens that were subjected to rotational fatigue, demonstrated different fracture behaviour. The first sign of damage accumulation was observed in the PFC which is the weaker component in these specimens (Figure 2.4A). The crack propagated from this region (Figure 2.4B) towards the interface between PFC and FRC (Figure 2.4C) resulting in debonding between the two layers. Afterwards the remaining fibres sustained the applied load until reaching their rupture strength (Figure 2.4D) leaving signs of brittle fracture on every single fibre.

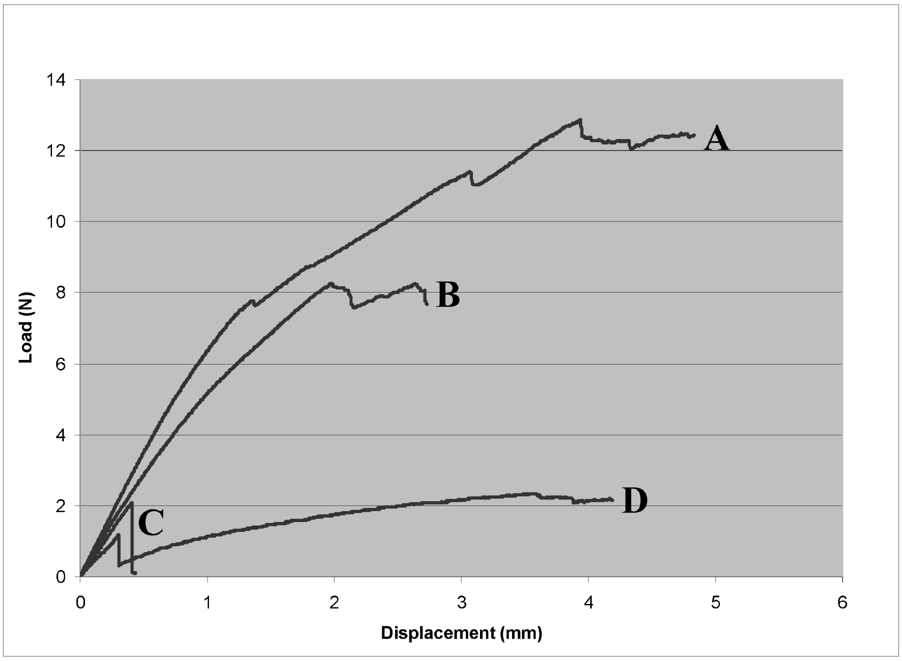

Figure 2.2 Load-displacement diagrams of the tested groups: PFC demonstrated elastic behaviour and sudden catastrophic failure (graph C). FRC demonstrated the highest fracture strength (graph A). FRC in compression (graph D) demonstrated initial sudden drop in load as soon as the PFC layer was broken. The highest toughness as associated when FRC was in tension (graph B) as demonstrated by the zigzag plateau following the highest point on the graph.

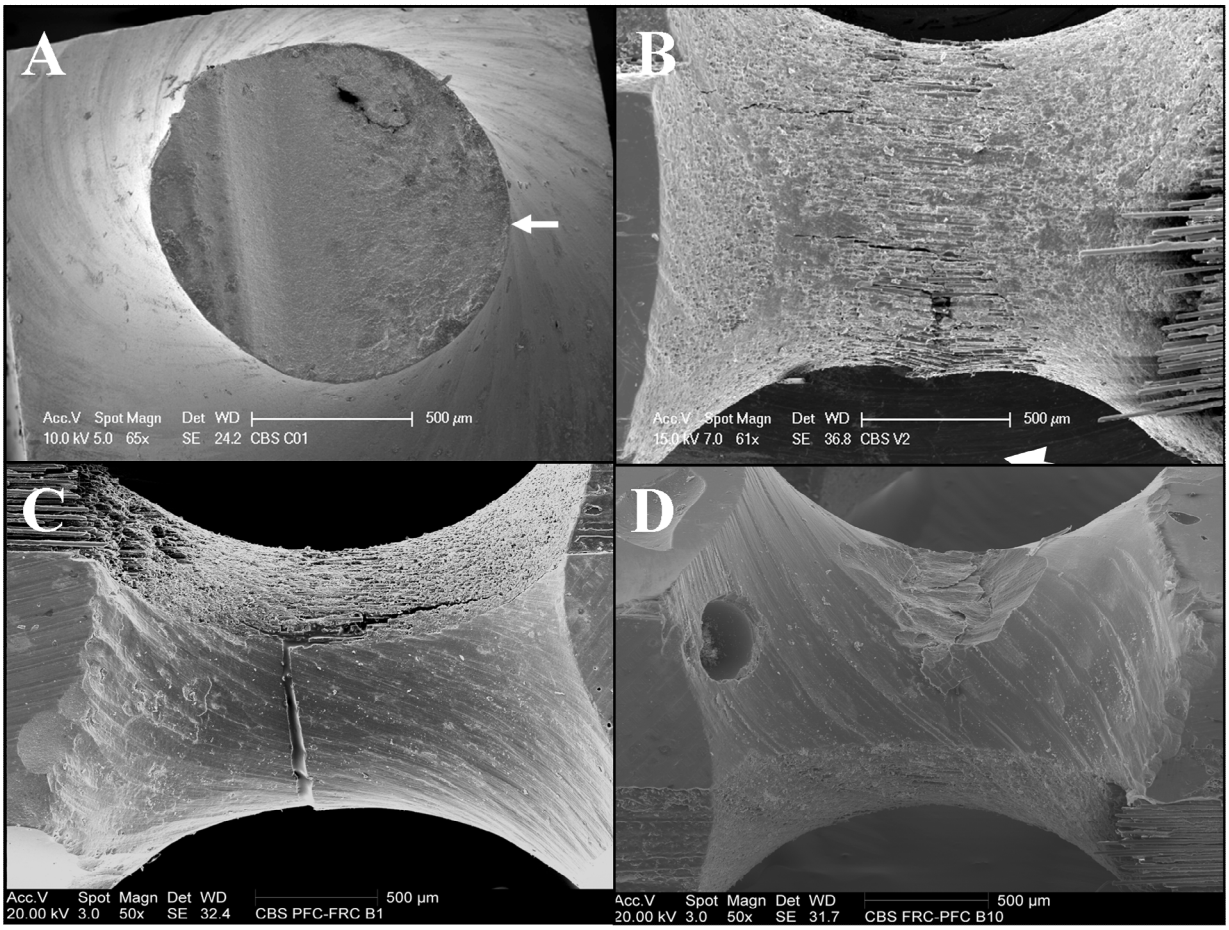

Figure 2.3 Scanning electron micrographs (SEM) of specimens subjected to a cantilever beam test: (A) SEM image, 65x, of a PFC specimen. Defect origin is demarcated with white arrow. Observe the compression curl on the opposite side. (B) SEM image, 61x, of a FRC specimen. Observe damage accumulation on the tensile surface meanwhile the compressive side remains intact. (C) SEM image, 50x, demonstrating deflection of the crack at the FRC-PFC interface of a FRC-compression specimen. (D) SEM image, 50x, of a FRC-tension specimen, demonstrating a compression crack

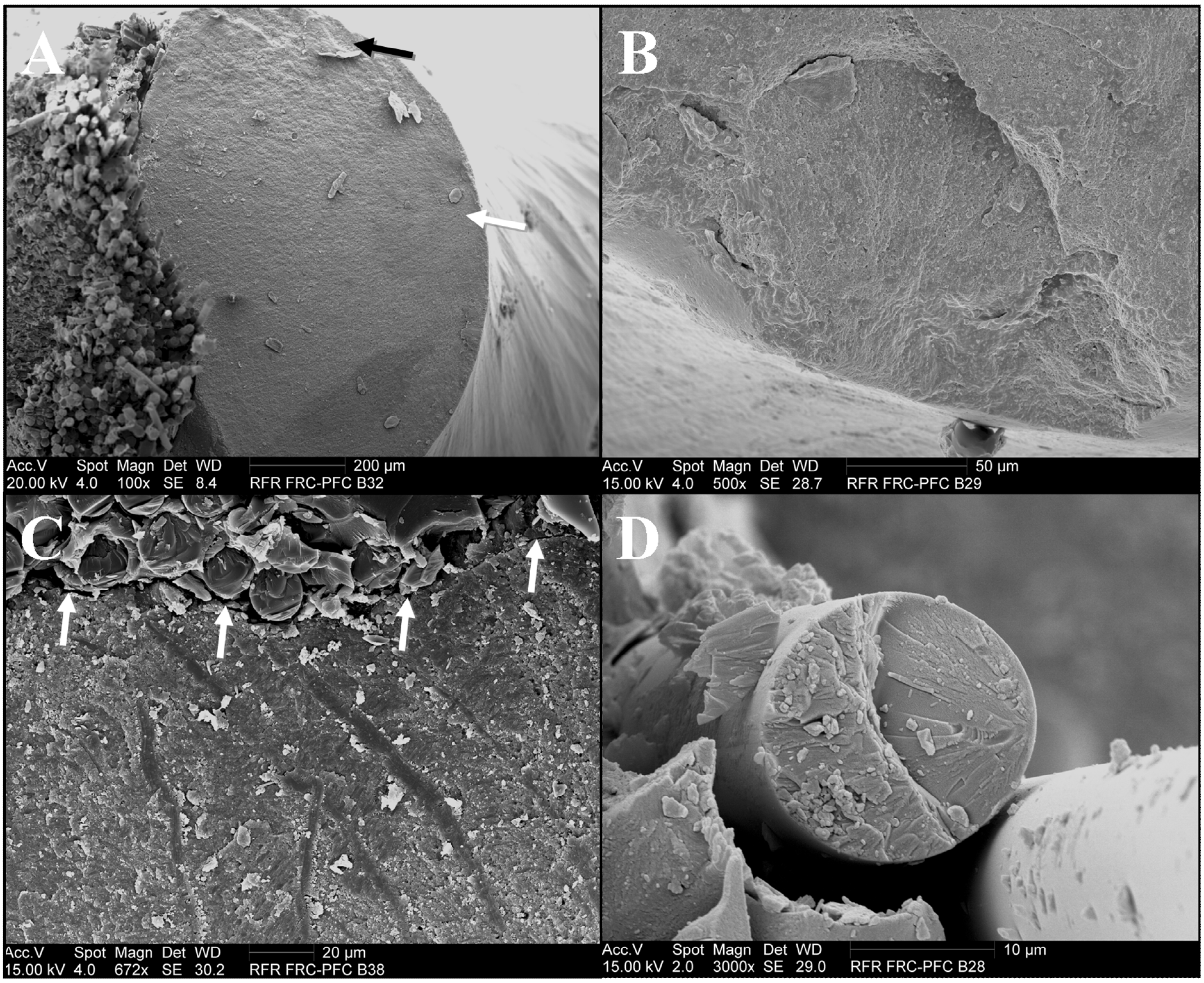

Figure 2.4 Scanning electron micrographs (SEM) of specimens subjected to a rotating cantilever beam fatigue test: (A) SEM image, 100x, of a bilayer specimen. Observe damage accumulation at the periphery of the specimen (black arrow) and defect origin (white arrow). (B) SEM image, 500x, detailed view of the defect origin of a bilayer specimen. (C) SEM image, 672x, demonstrating debonding at the FRC-PFC interface (white arrows) of a bilayer specimen. (D) SEM image, 3000x, demonstrating signs of brittle fracture on an independent fiber. Observe debonding between the fiber and the resin matrix.

2.5 Discussion

The resin-based composites used in this study are composed, on a structural level, of a mixture of different heterogeneous materials, which influences their mechanical properties. PFC on one hand is basically composed of brittle inorganic reinforcing filler particles with a high elastic modulus which are embedded in a quasi-brittle organic resin matrix. Nevertheless, it could be considered as single phase material with regard to measuring its mechanical properties as fracture strength, fatigue resistance, or work-of-fracture. On the other hand, the macroscopic size of the unidirectional fibre bundles used in FRC could influence its mechanical performance. This may offer a direct obstacle in the path of crack growth which requires paying careful attention to analyzing the behaviour of this material in order to obtain justified mechanical properties values. Finally, the bilayer of FRC and PFC should be considered as a two phase structure with consideration that the interface between these two different structures could also influence the performance of the specimens [24].

The rotational fatigue testing method applies a multi-vectorial stress on the specimens to be tested in a sequence of one tension and one compression per cycle. In fact, the direction of the applied stress represents the clinical situation where stresses on occlusal surfaces vary from parallel to the surface to perpendicular. The used test methodology subjects each point of the circumference of a specimen to tensile stress, implying calibration onto the weakest location [25]. According to Baran *et al.* [20] the staircase method virtually implies a fatigue limit and therefore not appropriate for lifetime predictions. It must be realized that the obtained test results do not represent the clinical fatigue life, but the method itself clearly indicates differences between and among the materials tested. The goal of the current investigation was rather to evaluate the differences in fatigue between the materials, than to predict the clinical fatigue. If this method not only reveals the mutual differences, but also predicts clinical behaviour, the question to be answered is which number of cycles represents the clinical situation. In the dental literature the opinions about the number of test cycles to be used, vary widely from 10^3 to 10^6 [26]. No hard evidence exists concerning the number of chewing strokes annually. Estimations have been made on these numbers per year and they vary considerably: Wiskott *et al.* [27] estimated 10^6 cycles annually. Huysmans *et al.* [28] concluded that composite restorations either fail before 10^4 cycles or after 10^5 cycles. Braem *et al.* [29] concluded that *in vitro* fatigue testing is not conclusive. Research into the relationship between stress and cycle numbers (S-N curves) [30] of unidirectional glass/epoxy composites revealed that low-cycle fatigue

(0.25 Hz) with high loads leads to failure in less than 10^4 cycles, supporting our choice of number of cycles. Furthermore, they observed a lower survival rate for low-cycle fatigue compared to high-cycle fatigue (5 – 10 Hz). The choice for the rotation frequency of 1.2 Hz has been made on a clinical basis, assuming that the upper limit of the chewing frequency is two strokes per second [29].

The outcome of fatigue experiments can be a relationship between stress and number of cycles, *i.e.* the S-N curve (endurance curve or Wöhler curve), or a fatigue resistance at a predetermined number of cycles. In comparison to the determination of the fatigue resistance the generation of a S-N curve involves a tedious and time consuming procedure. Determination of fatigue resistance at a predetermined number of cycles by the staircase method requires fewer tests (n = 15 - 20) and concentrates testing automatically near the mean.

Fiber-reinforcement significantly increased the fracture strength of resin-based composites. Flexural strength data found in the literature for Filtek Z100 range from 123 MPa to 151 MPa and are slightly lower than the 164.9 MPa found in this study [31,32]. Various factors might explain the slightly higher values found in our study. The experimental set-up might had influence on the results, so for that reason it should be noted that values from the literature were obtained by three-point bending tests instead of a cantilever beam test. Also the cross-sectional design and the L/D ratio (length/diameter) [33] of the specimens and the cross-sectional design, e.g. circular versus rectangular, can influence the obtained values. Flexural strength data for everStick obtained in this study (936.1 MPa) are comparable with the data found in literature (559 MPa till 1164 MPa) [34,35]. Only one study reports on the fracture strength of resin-based composites obtained by a cantilever beam test [36]. Although the exact values of both studies are incomparable, due to differences in specimen dimensions, fibre distribution and volume, a significant increase in fracture strength for FRC was noted in both studies [36]. When FRC was placed in compression, the fracture strength of the bilayer specimens was lower, but not significant different from homogeneous beams made of PFC resin. On the other hand, when the FRC was placed in tension, the strength was significantly increased, but still considerably below the fracture strength of homogeneous FRC beams (Table 2.2).

Few data are available on rotational fatigue resistance of dental resin-based composites. Only one study reports on the rotational fatigue resistance of PFC [37]. Scherrer *et al*. [37] studied the rotational fatigue resistance of PFC for provisional and definitive restorations and report a rotational fatigue resistance for the latter group (Artglass, Targis and Colombus) between 54.6 MPa and 62.1 MPa, which is comparable with the observed value of Filtek Z100 (51.5 MPa). When comparing the rotational fatigue resistance values to flexural strength values they observed a ratio between 38 and 62% which is higher than our value of 32%. A possible explanation is the difference in test set-up between fracture strength and fatigue resistance (3-point bending *vs*. cantilever beam), which is not the case in our study. Braem *et al*. [38] and Gladys *et al*. [39] reported only a difference of 30% between flexural strength and fatigue resistance of Filtek Z100. Although, it should be noted that they used a restrained 3-point bending set-up for the determination of fracture strength and fatigue resistance.

No literature is available on the rotational fatigue resistance of FRC. Nevertheless, the fatigue resistance of FRC is investigated in a multitude of studies [17-19,40,41]. Bae *et al*. [41] studied the dynamic fatigue strength of bar-shaped specimens made of a combination of FRC and PFC and tested in a three-point bending mode according the staircase method. Their fatigue strength values ranged from 90.2 MPa (Targis Dentine/Vectris Frame) and 196.9 MPa (Sculpture Body/FibreKor). In comparison our value of 116.5MPa obtained with specimens made of PFC and FRC is relatively low, which is obvious when taking the multi-vectorial nature of our test set-up into account. In the three-point bending test used by Bae *et al*. [41] FRC is subjected to tensile stresses and PFC to compressive stresses, while the rotating cantilever fatigue test subjects PFC as well as FRC to both types of stresses. Such a multi-vectorial stress application implies that the weakest material (PFC subjected to tensile stress) will cause failure. Although the incompatibility of the data we can conclude from the literature and also from our study that FRC are more fatigue resistant than PFC [20].

Work-of-fracture is the amount of energy needed to fracture a specimen and is a measure of toughness. It was clearly shown that specimens made of FRC (0.55 $kJ{\cdot}m^{-3}$) exhibit a higher work-of-fracture than specimens made of PFC (27.90 $kJ{\cdot}m^{-3}$) only. These results were in accordance with previous studies [42,43]. Petersen *et al*. showed that incorporation of glass fibres increased work-of-fracture [43] and showed the correlation between fibre volume fraction and work-of-fracture [42]. Specimens made of a combination of FRC and PFC, containing only half the amount of FRC and PFC, showed values in-between those of FRC and PFC.

Failure mode analysis of the broken specimens enabled better understanding of the failure mechanism of the different tested specimens. It is well known that the presence of fibres affects the fracture process which results in interrupting crack growth progression and thus enhances the fracture toughness of the FRC material.

Structural flaws are always present in the resin matrix and under the influence of cyclic loading micro cracks start to develop as the initial sign of failure. With continuous loading and due to the effect of stress concentration at these structural defects, micro cracks start to grow and join each other to form larger cracks serving as an entrance for oral fluids and bacteria, which may further accelerates the failure process. The presence of fibres ahead of and behind the crack tip significantly influences this process [24]. The presence of different components in one beam presented different physical barriers (including the influence of interface and bond strength between FRC and PFC resins) in the direction of crack propagation which resulted in preventing immediate failure and prolonged the failure process. This behaviour could be extremely beneficial in clinical conditions were fatigue is the most influential failure mechanism. Nevertheless, the restoration should be designed to bring the supporting fibres in tension in order to gain any strength benefit.

2.6 Conclusion

Although both cantilever beam test and rotating cantilever beam fatigue test are well established in the field of engineering, they are only introduced recently into the field of dentistry for the evaluation of resin-based composites. Within the limitations of this study, the following conclusions can be drawn: (i) the fatigue resistance of resin-based composites is lower than their fracture strength, (ii) FRC exhibits higher fracture strength and work-of-fracture than PFC, (iii) FRC are more fatigue resistant than PFC or combinations of FRC and PFC and, (iv) paying attention to the behaviour of fibre-reinforced composites is a key parameter to insure long term performance and adequate fatigue resistance.

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CHAPTER 3

Static and dynamic failure load of fibre-reinforced composite and particulate filler composite cantilever resin-bonded fixed dental prostheses

3.1 Abstract

Objectives: The aim of this study was to evaluate in vitro the influence of fibre-reinforcement and luting cement on the static failure load (SFL) and dynamic failure load (DFL) of simulated two-unit cantilever resin-bonded fixed dental prostheses (RB-FDPs).

Materials and Methods: Forty-six particulate filler composite (PFC) beams and seventy-six fibre-reinforced composite (FRC) beams were prefabricated and subsequently luted (RelyX ARC or Panavia F2.0) onto flat ground bovine enamel. The SFL of the different specimen types was determined with a peel test and the DFL was determined with a rotating cantilever beam fatigue testing device.

Results: The PFC specimens showed a significantly lower SFL than the FRC specimens. The luting cement showed a significant effect on the SFL of the PFC specimens, but not with FRC. The DFL of PFC specimens was significantly lower than for FRC specimens. The luting cement showed a significant effect on the DFL of the PFC specimens, but not so with FRC. With both the SFL and the DFL tests all PFC beams fractured, leaving the bonded part on the tooth surface, but FRC beams partially debonded from the tooth surface, leaving to a varying extent fibres connected to the enamel surface. Coincidentally the uncured fibres turned out to be prone to aging, which effect has been investigated.

Conclusions: Within the limitations of this study it can be concluded that PFC without fibre reinforcement is not suitable for the fabrication of two-unit cantilever RB-FDPs, despite the significant effect of the luting cement, but FRC is suitable.

3.2 Introduction

Minimally invasive dentistry became an important aspect of contemporary dentistry [1]. Not only the branch of cariology and operative dentistry, but also that of prosthetic dentistry adopted the concept of tooth tissue preservation. In the field of prosthodontics this paradigm shift can be noticed by the regained interest for resin-bonded fixed dental prostheses (RB-FDPs). The development of adhesive dentistry and modified preparations, specifically designed for the materials used, have made RB-FDPs a viable and reliable treatment for the replacement of a missing tooth [2]. Two-unit cantilever RB-FDPs came into focus after the observation that often just one of the retainers of their three-unit fixed-fixed counterparts debonded. Many of these partially debonded RB-FDPs were successfully converted into a cantilever design after removal of the debonded retainer. Elimination of interfacial stresses, induced by dynamic tooth contacts and differential movements of the abutment teeth, provides a rationale for introducing two-unit cantilever RB-FDPs in clinical practice [3,4]. Several clinical studies of the last decade demonstrated the reliability of two-unit cantilever RB-FDPs [5-8].

Besides the concept of minimal invasive dentistry the variety of materials available for dentistry increased during the last decades. This resulted in new applications and designs of metal-free restorations, *e.g.* all-ceramics and fibre-reinforced composites. The increasing popularity of these metal-free restorations can be attributed to two major reasons. The first reason is that the community is becoming more aware of the possible adverse health effects of base alloys used in dentistry [9]. Secondly, patients are not only seeking dental treatment for reasons of pain or functional discomfort, but also because of aesthetic concerns. For particulate filler composite and glass fibre-reinforced composite, dental literature only provides suspicions of adverse health effects for patients [10] concerning the resin part, but no health effects of glass fibres are known so far. Occupational health hazards from dental composites for dentists and their personnel are known for quite some time now [10,11]. One of the major concerns of fibre-reinforced composites is that there is less evidence about the survival rate compared to metal, porcelain fused to metal, and all-ceramic restorations.

The survival rate of dental restorations is not only compromised by high static loads, but also by low cyclic loads, the latter known as fatigue. Fatigue is the phenomenon where failure is induced by subjecting the material or structure to repeated sub-critical loads [12]. Mechanical failure of dental restorations can be attributed to fatigue in most of the cases. *In vitro* fatigue studies can be designed as, for example, repeated loading 3-point or 4-point bending, tensile or compressive strength tests. An elegant way of measuring fatigue is by a rotational cantilever beam test. This fatigue testing method was introduced in dentistry in the early 1990's by Wiskott and co-workers [13] and it has been used in several studies on soldered joints [13,14], bond strength [15], resin-based composites [16], post types [17], implantology [18,19]. The outcome of these experiments can be a relationship between stress and number of cycles, *i.e*. the S-N curve (endurance curve or Wöhler curve), or a fatigue resistance at a predetermined number of cycles. It has to be mentioned that opposite to the determination of the fatigue resistance the generation of a S-N curve involves a tedious and time consuming procedure.

The aim of this study was to evaluate *in vitro* the influence of fibre-reinforcement and luting cement on the static failure load (SFL) and dynamic failure load (DFL) of two-unit cantilever resin-bonded fixed dental prostheses by using simplified cantilever beams as depicted in Figure 3.1. The static and dynamic failure loads of particulate filler composite (PFC) and fibre-reinforced composite (FRC) beams luted with two different cements were evaluated. Also the effect of shelf-life of fibres will be evaluated since in common practice opened packages will be stored and used after some time.

3.3 Materials and Methods

Two resin-based composites, one particulate filler composite (Filtek Z100, shade A2, 3M Espe, St Paul, MN, USA) and one fibre-reinforced composite (everStick C&B, Sticktech Ltd, Turku, Finland), were selected for the fabrication of simulated two-unit cantilever RB-FDPs. The compositions of the materials used in this study are summarized in Table 3.1.

Table 3.1 Materials used in the study.

Brand	Composition	Manufacturer	Lot number
Filtek Z100	Resin: Bis-GMA, TEGDMA; Filler: zirconia, silica (≈ 66 vol%)	3M-ESPE, St Paul, MN, USA	20061003
everStick C&B	Resin: PMMA, Bis-GMA; Filler: silanised E-glass fibres (≈ 65 vol%)	Sticktech Ltd., Turku, Finland	2061010-ES-165
Stick resin	Bis-GMA, TEGDMA	Sticktech Ltd., Turku, Finland	550 9986
Panavia F2.0	EDII primer and luting resin	Kuraray medical Inc, Okayama, Japan	41170
ED II Primer	Primer A: HEMA, MDP, 5-NMSA, water, accelerator Primer B: 5-NMSA, accelerator, water, sodium benzene sulphinate		
Luting resin	Base paste: hydrophobic aromatic (and aliphatic) dimethacrylate, hydrophilic dimethacrylate, sodium aromatic sulfinate, N,N-diethanol-p-toluidine, functionalized sodium fluoride, silanized barium glass Catalyst paste: MDP, hydrophobic aromatic (and aliphatic) dimethacrylate, hydrophilic dimethacrylate, silanized silica, photoinitator, dibenzoyl peroxide		
Clearfil Porcelain Bond Activator	Hydrophobic dimethacrylate, MPTS, Bis-PMA	Kuraray medical Inc, Okayama, Japan	00158B
Clearfil SE Bond Primer	MDP, HEMA, hydrophilic dimethacrylate, dl-camphorquinone, N,N-diethanol-p-toluidine water	Kuraray medical Inc, Okayama, Japan	00407A
RelyX ARC	Scotchbond 1 and luting resin	3M-ESPE, St Paul, MN, USA	20040309
Scotchbond 1	Bis-GMA, HEMA, dimethacrylates, polyalchenoic acid, copolymer, ethanol, water 3–8%, initiators		
Luting resin	Paste A: Bis-GMA, TEGDMA, zircon/silica filler (68 wt%), photoinitiators, amine, pigments Paste B: Bis-GMA, TEGDMA, benzoic peroxide, zircon/silica filler (67 wt%)		

Bis-GMA bisphenol-A-glycidyl dimethacrylate; TEGDMA triethylenglycol dimethacrylate; MDP 10-methacryloyloxydecyl dihydrogen phosphate; HEMA 2-hydroxyethyl methacrylate; 5-NMSA N-methacrylyloyl 5-aminosalicylic acid; MPTS 3-methhacryloxypropyl trimethoxy silane; Bis-PMA bisphenol-A-polyethoxy dimethacrylate

Specimen preparation

The buccal surfaces of bovine teeth were flat ground with 600 grit SiC-paper on a grinder/polisher (Ecomet, Buehler Ltd, Lake Bluff, IL, USA) in a way dentine was not exposed. Bar specimens (2.6 x 2.7 x 10.0 mm) were cut, using a slow-speed water-cooled diamond saw (Isomet 1000, Buehler Ltd, Lake Bluff, IL, USA). The specimens were stored in tap water at 5ºC until use without adding any antimicrobial agent.

Forty-six beams (1.0 x 2.0 x 25.0 mm) of particulate filler composite (Filtek Z100, shade A2, 3M Espe, St Paul, MN, USA) and seventy-six fibre-reinforced composite beams (1.0 x 2.0 x 25.0 mm) (everStick C&B, Sticktech Ltd, Turku, Finland) were made using a custom-made mould. Of these seventy-six FRC beams forty-six were made of fibres from a freshly opened package (fresh fibres), while the other thirty beams were made of fibres from a package that had been opened six months before (aged fibres) and subsequently stored according to the manufacturer's instructions. Specimens were light-cured for 40 s by a handheld polymerisation unit (Elipar Highlight, 3M Espe, St Paul, MN, USA) with a power out-put of 800 $mW \cdot cm^{-2}$ (Curing Radiometer model 100, Demetron Research Corporation, Danbury, USA).

All PFC and FRC beams were immediately after preparation luted onto the enamel surface of the bovine teeth bars (bonding surface 2.0 x 5.0 mm^2) with either one of two commercially available resin luting cements and their proprietary adhesive according to the manufacturer's instructions (RelyX ARC with Scotchbond 1, 3M ESPE, St Paul, MN, USA and Panavia F2.0 with ED primer, Kuraray Medical Inc, Okayama, Japan). Twenty-three PFC beams were luted with RelyX ARC, while the other twenty-three PFC beams were luted with Panavia F2.0. Twenty-three FRC beams with new fibres and fifteen FRC beams with aged fibres were luted with RelyX ARC. Another twenty-three FRC beams with fresh fibres and fifteen FRC beams with aged fibres were luted with Panavia F2.0. Surface conditioning methods of the beams differed according to the material:

(i) PFC beams were sandblasted (Vaniman, Fallbrook, CA, USA) with 50 μm alumina particles (Korox 50, Bego, Bremen, Germany) under 0.3 MPa pressure for 3 s followed by cleaning with compressed air for 5 s. Subsequently, the adhesive surface was silanised by applying an equal mixture of Clearfil SE Bond Primer and Clearfil Porcelain Bond Activator (Kuraray medical Inc, Okayama, Japan) for 30 s.

(ii) The IPN-matrix of the FRC beams was reactivated by applying a thin layer of resin (Stick Resin, Sticktech Ltd, Turku, Finland) for 5 minutes [20].

During cementation the beams were kept under a constant pressure of 50 N for 60 s, in order to obtain a cement layer with uniform thickness, before all edges were light cured for 20 s. All specimens were water stored at 37ºC for 72 h before testing.

Static failure load

Static failure load of simulated two-unit cantilever RB-FDPs was measured according to a peel test, demonstrated by Van Dalen *et al*. [21] and, within the context of their research, considered to be the most relevant test simulating the clinical situation (Figure 3.1). The load at the cantilever beams was applied at a distance of 10 mm from the bovine teeth. Specimens (n = 8) were loaded until failure in a universal testing machine (Hounsfield 12B AD, model 20-30, Salfords, UK) at a crosshead speed of 1 mm·min^{-1}. All fractured specimens were visually examined and their mode of failure was recorded.

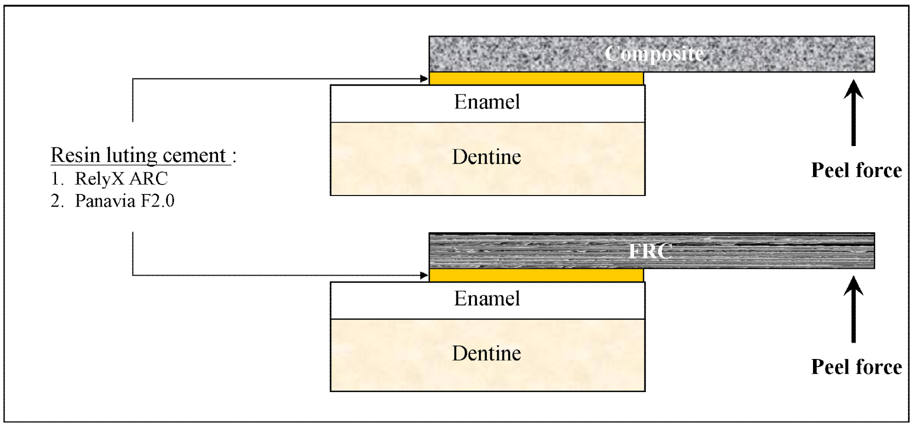

Figure 3.1 Test arrangement of the experimental set-up for the determination of static failure load.

Dynamic failure load

The dynamic failure load of simulated two-unit cantilever RB-FDPs was determined by using a rotating cantilever beam fatigue testing device (Figure 3.2) according to the staircase approach. With this method specimens (n = 15) are tested for a chosen number of cycles, assuming that fatigue occurs at a lower stress level than the previously determined static failure load. According to the principles of the staircase approach [22] the initial test stress was set at 50% of the previously determined static failure load.

The specimens were mounted in the chuck of the fatigue device in a way the adhesive interface had an exact centric alignment. A ball-bearing was fixed to the beams at a distance of 10 mm from the luting interface. The stress at the resin luting interface was induced by attaching a weight onto the ball-bearing. The specimens were rotated at 1.2 Hz for 10^4 cycles or until failure. If failure occurred before 10^4 cycles the stress was decreased with 10% of the original static failure load, respectively increased with the same percentage when the specimen survived [23]. The dynamic failure load was investigated both with a freshly opened package of fibres (fresh FRC) and with fibres which were stored at 4ºC in a pre-opened package for 6 months in the dark (aged FRC).

All fractured specimens were visually examined and their mode of failure was recorded if possible. When visual inspection yielded insufficient information, SEM images were obtained and interpreted.

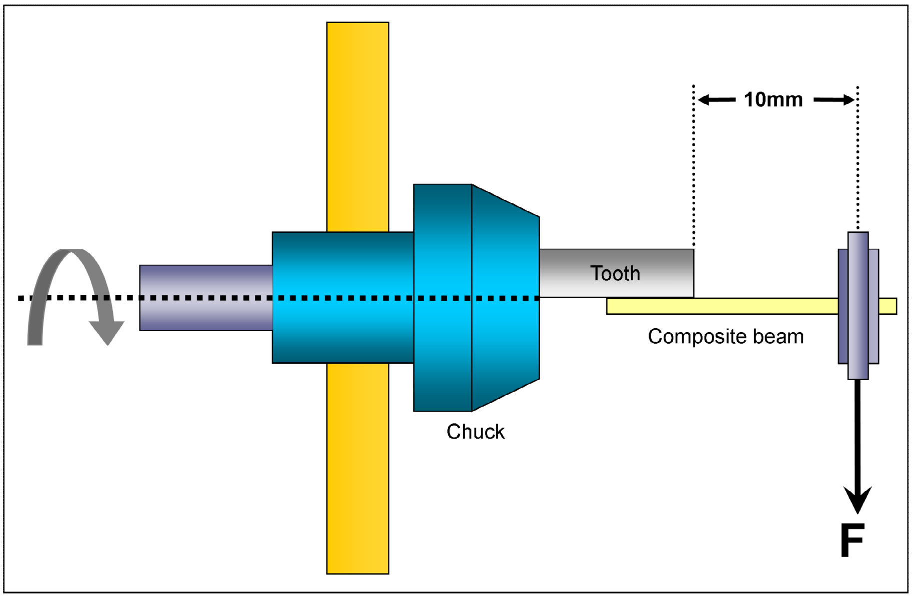

Figure 3.2 Schematic representation of the working principle of the rotating cantilever beam fatigue testing device: the simulated cantilever RB-FDPs are rotating around an axis which coincides with the longitudinal centre line of the cement layer while the composite beams are stressed by attaching a weight (*F*) to a ball-bearing.

Statistical Analysis

Statistical analysis was performed with the statistical software SigmaStat 3.0 (SPSS Inc. Chicago, IL, USA). The data obtained from the fatigue test were analysed using multiple logistic regression in order to determine the mean dynamic failure load and standard deviation. The dynamic failure load can be defined as the load at which 50% of the specimens fail.

Mean and standard deviations of static failure loads for each group were calculated from the dataset generated by the static loading experiment (peel test).

The obtained means and standard deviations were compared based on the mean, standard deviation and group size. This type of evaluation was performed earlier by Yoshida K *et al*. [24]. One-way analysis of variance (ANOVA) followed by Tukey's post hoc test was performed to determine the effect of luting cement and material on the observed static and dynamic failure loads. P-values of less than 0.05 were considered to be statistically significant.

3.4 Results

The static and dynamic failure loads of each material and resin luting cement combination and the DFL/SFL ratio are summarised in Table 3.2.

Table 3.2 The mean SFL and DFL, with SD in parentheses, of simulated two-unit cantilever RB-FDPs with different luting cements and materials. Test groups with the same superscript letter are not statistically different.

	Static failure load (N)	Dynamic failure load (N)		Ratio	
		fresh FRC	aged FRC	fresh FRC	aged FRC
Z100 - Panavia F2.0	2.46^{a}(0.28)	0.60^{d}(0.08)		0.24	
Z100 – RelyX ARC	3.76^{b}(0.23)	0.84^{e}(0.11)		0.23	
everStick – Panavia F2.0	5.19^{c}(1.12)	2.12^{f}(0.02)	1.40^{g}(0.09)	0.41	0.27
everStick – RelyX ARC	5.22^{c}(1.16)	2.09^{f}(0.02)	1.46^{g}(0.03)	0.40	0.28

Static failure load

One-way ANOVA showed that the static failure load of specimens made of PFC was significantly lower ($F = 20.4$; $p < 0.001$) than the static failure load of FRC specimens. Luting cement had significant effect ($p < 0.05$) on the static failure load of PFC specimens, but no significant effect ($p > 0.05$) on static failure load of FRC specimens was recorded.

PFC specimens failed due to fracture of the pontic part of the beam, in a way that the bonded part was left on the tooth surface. On the other hand, all FRC specimens partially debonded *i.e.* debonded cohesively within the fibre bundle, from the tooth surface, leaving to a varying extent fibres connected to the enamel surface, actually creating a loose fibre connection between the beam and the enamel surface (Figure 3.3).

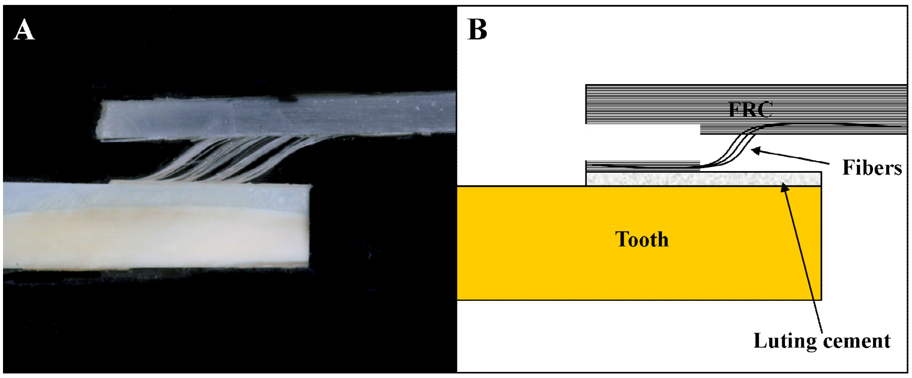

Figure 3.3 Failure mode of FRC beams: loose fibre connection between the beam and the enamel surface after failure. Photograph (A) and schematic representation (B) of a partially debonded FRC beam.

Dynamic failure load

The results of the staircase method represented as an up-and-down graph are depicted in Figure 3.4. Logistic regression analysis of PFC specimens, fresh and aged FRC specimens luted with different resin luting cements are graphically represented in Figure 3.5. One-way ANOVA showed that the dynamic failure load of PFC specimens was significantly lower than the dynamic failure load of FRC specimens. Luting cement had no significant effect ($p > 0.05$) on dynamic failure load of FRC specimens. The dynamic failure load of the fresh FRC was significantly higher than that of the aged FRC.

The failure mode of PFC specimens after fatigue loading was comparable with those after static loading: fracture occurred in the pontic part, while the bonded part was left on the tooth surface. Fatigue failure of FRC specimens presented itself as debonding.

SEM inspection after fatigue of the fibre surfaces of both the Panavia (Figure 3.6A) and the RelyX ARC (Figure 3.6C) luted specimens reveals that the matrix has debonded from the fibres, leaving the fibres exposed. Actually, the bond between the matrix and both luting cements proved stronger than the bond between matrix and fibres. Consequently, SEM inspection of the enamel surfaces of both specimen types reveals the matrix which was missing from the fibre bundles left behind on the enamel surface (Figure 3.6B and 3.6D). When the fibre bundle is considered as a homogeneous material, the failure is cohesive within this bundle. But when considering this phenomenon as a failure between matrix and fibres, it is an adhesive failure.

The DFL/SFL ratio as depicted in Table 3.2 reveals that both PFC ratios are almost equal as are both FRC ratios. This result excludes any luting cement influence.

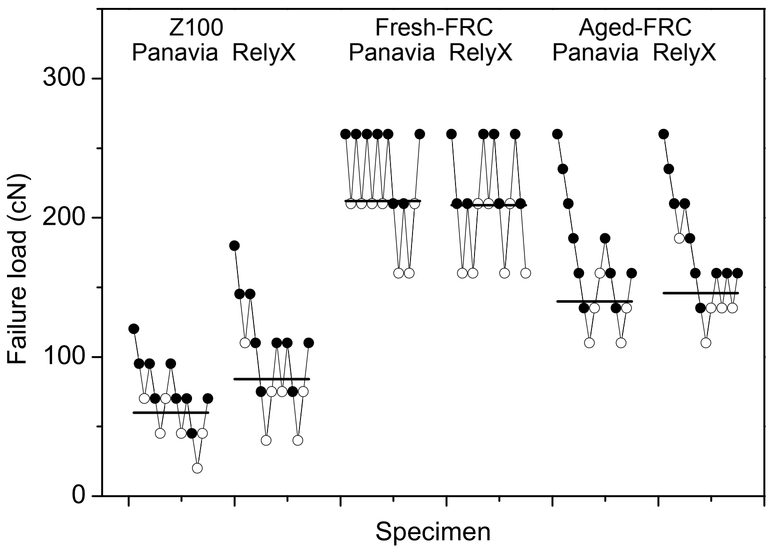

Figure 3.4 Graphical representation (up-and-down graph) of the results of the staircase method after 10^4 cycles for all groups. Open symbols represent no failure, while solid symbols represent failure.

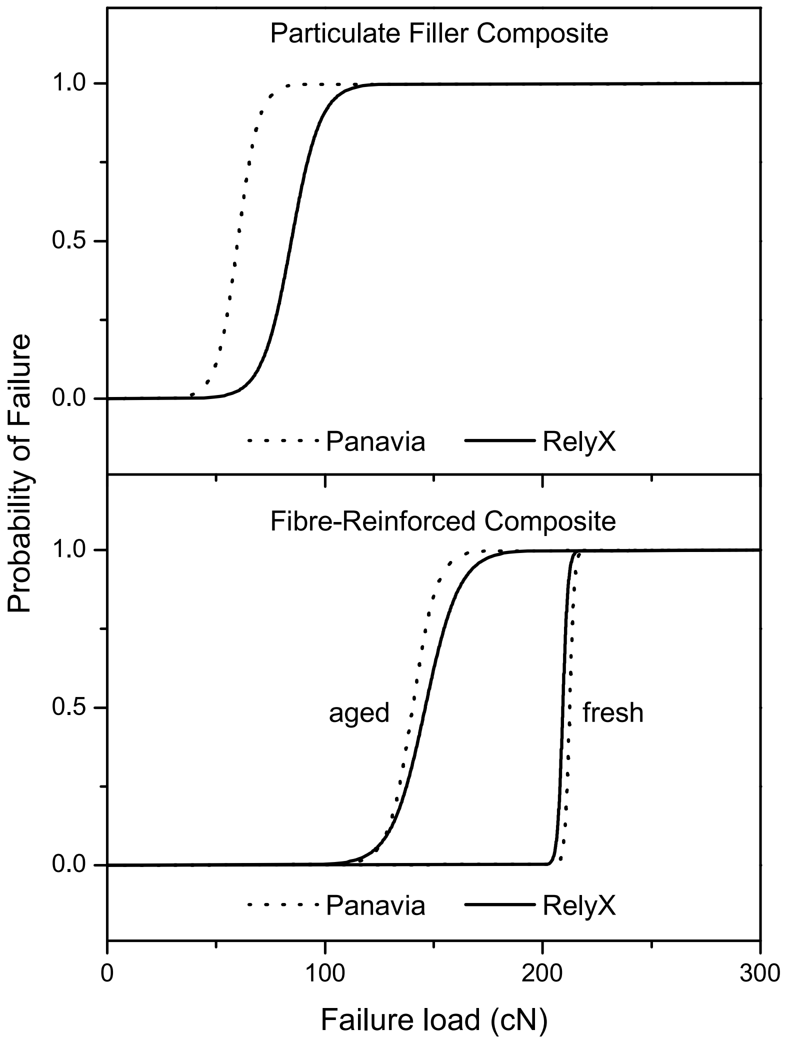

Figure 3.5 Graphical representation of fatigue data: (top) PFC beams and (bottom) FRC beams. The logistic regression curve is showing the probability of failure at each load level for each combination.

3.5 Discussion

The static failure load of a simulated cantilever RB-FDP can be determined in different ways. It was shown that a peel test generated the lowest failure loads compared to a load test and a torque test, and therefore can be considered to be clinically the most frequent and relevant failure mechanism [21,25]. Also in the present study the static failure load of simulated cantilever RB-FDPs was determined in a laboratory model where tensile peel stresses were generated. Cantilever RB-FDPs made of PFC showed both the highest static and dynamic failure load when combined with a low E-modulus resin luting cement (Table 3.2). RelyX ARC has an E-modulus of 5.6 GPa [26], while the Panavia E-modulus is 12.8 GPa [27,28]. The effect of the low E-modulus is a confirmation of earlier findings of Van Dalen *et al*. [21]. It was assumed that luting cements with a lower E-modulus allow a more even stress distribution within the cement layer, leading to lower peak stresses and a higher load to failure, *i.e*. static failure load. Contrary to these results there is no significant difference between the FRC-RelyX ARC and the FRC-Panavia combinations, respectively. The explanation for this phenomenon is that the bond between matrix and luting cement is stronger than the bond between fibres and matrix (Figure 3.3). The latter will therefore fail prematurely, which is confirmed by the SEM images (Figure 3.6).

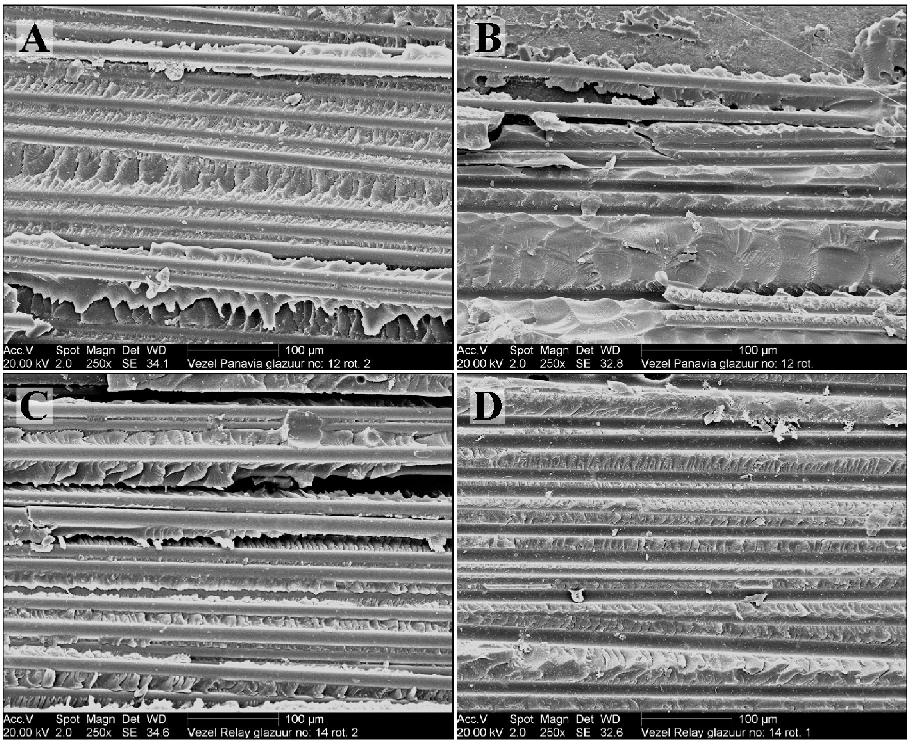

Figure 3.6 SEM images (magnification 250x) of the fracture surfaces of simulated cantilever resin-bonded FRC-FDPs luted with Panavia F2.0 (A: fibre surface; B: enamel surface) and RelyX ARC (C: fibre surface; D: enamel surface) after fatigue with B and D clearly showing fibres which delaminated from the matrix.

The rotational fatigue testing method provides a multi-vectorial stress on the specimens to be tested in a sequence of one tension and one compression per cycle. In fact, the direction of the applied stress represents the clinical situation where stresses on occlusal surfaces vary from parallel to the surface to perpendicular. To what extent the obtained test results are predictive for the clinical situation remains to be considered [29], but the method itself clearly indicates differences between and among the materials tested. If this method not only reveals the mutual differences, but also predicts clinical behaviour, the question to be answered is which number of cycles represents the clinical situation. In the dental literature the opinions about the number of test cycles to be used, vary widely from 10^3 to 10^6 [15]. No hard evidence exists concerning the number of chewing strokes annually. Estimations have been made on these numbers per year and they vary considerably. Wiskott *et al*. [14] estimated 10^6 cycles annually. Huysmans [30] concluded that composite restorations either fail before 10^4 cycles or after 10^5 cycles. Braem *et al*. [31] concluded that *in vitro* fatigue testing is not conclusive. Research into the relationship between stress and cycle numbers (S-N curves) [32] of unidirectional glass/epoxy composites revealed that low-cycle fatigue (0.25 Hz) with high loads leads to failure in less than 10^4 cycles, supporting our choice of number of cycles. Furthermore, they observed a lower survival rate for low-cycle fatigue compared to high-cycle fatigue (5 – 10 Hz). The choice for the rotation frequency of 1.2 Hz has been made on a clinical basis, assuming that the upper limit of the chewing frequency is two strokes per second [31].

Aging (water storage and mechanical loading) of polymerized FRC affects their mechanical properties in a negative way, by reducing their strength by almost 30% [33]. An active aging process of the specimens has not been executed in this study. But leaving passively a part of the fibre packages open for six months before use, lead to a drop in dynamic failure load of more than 30%. Therefore, the present study strongly indicates that short-term storage of pre-opened, non-polymerized FRC induces a form of passive aging. A similar effect with the PFC has never been described in the literature, most likely for the simple reason that PFC compules can not be pre-opened. Factors influencing the mechanical properties of a FRC are volume, orientation and location of the fibres, and the quality of the chemical bond between the components [34,35]. It has been shown that loss of interfacial bond between fibres and matrix is the primary cause of reduction in mechanical properties [33]. FRCs, being exposed to the oral environment and, as in the present study, exposed in a pre-opened package to an environment with a certain extent of humidity [36], are subjected to watersorption, which causes a small increase in volume. Watersorption is a mechanism of water penetrating into the resin matrix itself [37,38] and also between matrix and fibres, due to the existence of voids along the fibres, caused by incomplete fibre-matrix impregnation [36,38,39]. Watersorption induces plasticisation of the resin matrix and deteriorates the fibre-matrix interphase by possible leaching of glass forming oxides from the fibre surface and by hydrolytic degradation of the polysiloxane network between fibres and matrix [38,40]. This process weakens the structure, inevitably leading to a decrease in dynamic failure load (Table 3.2). It has to be noticed that semi-IPN matrix-based FRCs, as used in the present study, in comparison to UTMA-matrix-based FRCs [38] or PFC [41] are more prone to watersorption, with as a possible explanation the difference in filler content [41] and hydrophilic properties of the resin matrix [38]. The effect of long-term storage of pre-opened packages FRC on their mechanical properties is an interesting topic for further research.

3.6 Conclusion

Within the limitations of this study we can conclude that:

1. The dynamic failure load of simulated two-unit cantilever RB-FDPs is lower than their static failure load.
2. FRC beams in comparison to PFC beams generate higher static and dynamic failure loads.
3. Aged fibres have a lower dynamic failure load than fresh fibres.

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CHAPTER 4

Three-dimensional finite element analysis of anterior two-unit cantilever resin-bonded fixed dental prostheses

4.1 Abstract

Objectives: The aim of this study was to evaluate, by finite element analysis (FEA), the influence of different framework materials on the biomechanical behaviour of anterior two-unit cantilever resin-bonded fixed dental prostheses (RB-FDPs).
Materials and Methods: The 3D FEA model consisted of a two-unit cantilever RB-FDP replacing a maxillary lateral incisor with a wing-shaped retainer on the central incisor and an adjacent canine. Five different framework materials were compared: direct fibre-reinforced composite (FRC-Z250), laboratory fibre-reinforced composite (FRC-ES), metal (M), glass-ceramic (GC) and zirconia (ZI). The isotropic materials were veneered with isotropic feldspathic porcelain, while the anisotropic material was veneered with isotropic particulate filler composite. A stress of 90 MPa at a 45° angle was applied to the incisal edge of the pontic.
Results: A similar stress pattern, with tensile stresses in the connector area, was observed in RB-FDPs for all materials. Maximal principal stress showed a decreasing order: ZI (239.6 MPa) > M (197.1 MPa) > GC (178.4 MPa) > FRC-ES (177.1 MPa) > FRC-Z250 (156.9 MPa). The maximum displacement of RB-FDPs was higher for FRC-Z250 (0.048 mm) and FRC-ES (0.035 mm) than for M (0.019 mm), GC (0.019 mm) and ZI (0.017 mm). Stress analysis depicted differences in location of the maximum stress at the luting cement interface between materials. For FRC-Z250 and FRC-ES the maximum stress was located in the upper part of the proximal area of the retainer, whereas for M, GC and ZI the maximum stress was located at the cervical outline of the retainer.
Conclusions: Within the limitations of this study, FEA revealed differences in biomechanical behaviour between RB-FDPs made of different framework materials. The general observation was that a RB-FDP made of FRC provided a more favourable stress distribution.

4.2 Introduction

Resin-bonded fixed dental prostheses (RB-FDPs) have proven to be a reliable treatment alternative for the replacement of missing teeth [1] especially in cases were conservation of tooth tissue is needed and limited financial resources are available. According to a recent systematic review, RB-FDPs exhibit an estimated survival rate of 87.7% (95% confidence interval: 81.6%-91.9%) after 5 years [2]. Notwithstanding their good clinical performance, the most frequent complication was debonding, which occurred in 19.2% (95% CI: 13.8-26.3%) of RB-FDPs over an observation period of 5 years [2].

The use of more extensive preparation of the abutment teeth, including palatal or lingual coverage with 180-degree wrap-around, chamfer, cingulum rests, and proximal guide planes and grooves, is a way to improve the retention of RB-FDPs [3]. Another way to minimize debonding is to design RB-FDPs as a two-unit cantilever. This approach came into focus after the observation that many partially debonded three-unit fixed-fixed RB-FDPs could be successfully converted into a two-unit cantilever design after removal of the debonded retainer [4]. Elimination of interfacial stresses, induced by a combination of dynamic tooth contacts and differential movements of the abutment teeth, is the most widely accepted explanation for their successful clinical performance [3,5]. Several clinical studies of the last decade have demonstrated that two-unit cantilever RB-FDPs performed as well as or even better as their three-unit fixed-fixed counterparts [4,6-10].

The framework of RB-FDPs is traditionally made of metal alloys, but their poor aesthetics and the growing awareness towards possible adverse health effects of dental alloys [11] stimulated the interest in metal-free restorations. Nowadays, all-ceramics [10] and fibre-reinforced composites (FRC) [12] are viable alternatives for framework fabrication of RB-FDPs. Some clinical cases reported promising results for all-ceramic RB-FDPs [13,14]. In addition Kern *et al*. reported 5-year survival rates of 73.9 % for three-unit fixed-fixed designs and 92.3% for two-unit cantilever designs [10]. A recently published systematic review reported for FRC-FDPs a survival rate of 73.4% (95% CI: 69.4-77.4%) after 4.5 year [15]. During a 5 year multicenter clinical study FRC RB-FDPs exhibited a survival rate of 64% [16]. The differences in material properties, especially elastic modulus, adhesive properties and thermal expansion coefficient are believed to affect the mechanical and clinical performance of RB-FDPs [17]. In order to better understand the failure mechanism of two-unit cantilever RB-

FDPs, increased knowledge on the biomechanical behaviour of these restorations is needed.

The aim of the present study was to compare, by means of three-dimensional finite element analysis (3D FEA), the biomechanical behaviour of anterior two-unit cantilever RB-FDPs made of various framework materials.

4.3 Material and Methods

Definition of structures, geometric conditions, and materials

In order to create a FE model, a physical model of a single tooth gap in the anterior right maxilla, consisting of a central incisor, a missing lateral incisor and a canine (Figure 4.1A), was created. The central incisor served as the abutment tooth, but was not provided with a retainer preparation. The missing lateral incisor was replaced by a two-unit cantilever RB-FDP (Figure 4.1C) with a retainer on the central incisor. A wing-shaped retainer design, which enwrapped the palatal and distal surface of the abutment tooth, was selected and the pontic was shaped according a modified ridge lap design.

A dental CAD/CAM system (Dental Cadim 107D, Advance Co. Ltd., Tokyo, Japan) was used for measuring the model of the single tooth gap and the replica of the FDP at 0.25mm intervals, where after the captured data points were plotted in a 3D CAD software (VX 7.5, VX Co. Ltd., Florida, USA) in order to construct the 3D model. The 3D model of the single tooth gap and the RB-FDP were joined together and subsequently the cement layer was created manually. The model was converted to 3D solid models (ANSYS 11 Sp, ANSYS Inc., Houston, TX, USA).

The geometry of the healthy standard tooth as abutment has been previously described [18]. Not only the natural tooth geometry, but also the composition was mimicked, by including enamel, dentin and pulp tissues into the models. On the basis of the contours of the solid model, root under the bone, periodontal ligaments and alveolar bone volumes were not created. Three-dimensional FE model of the cement layer is shown in Figure 4.1B. The thickness of the cement layer was maintained at 100 μm.

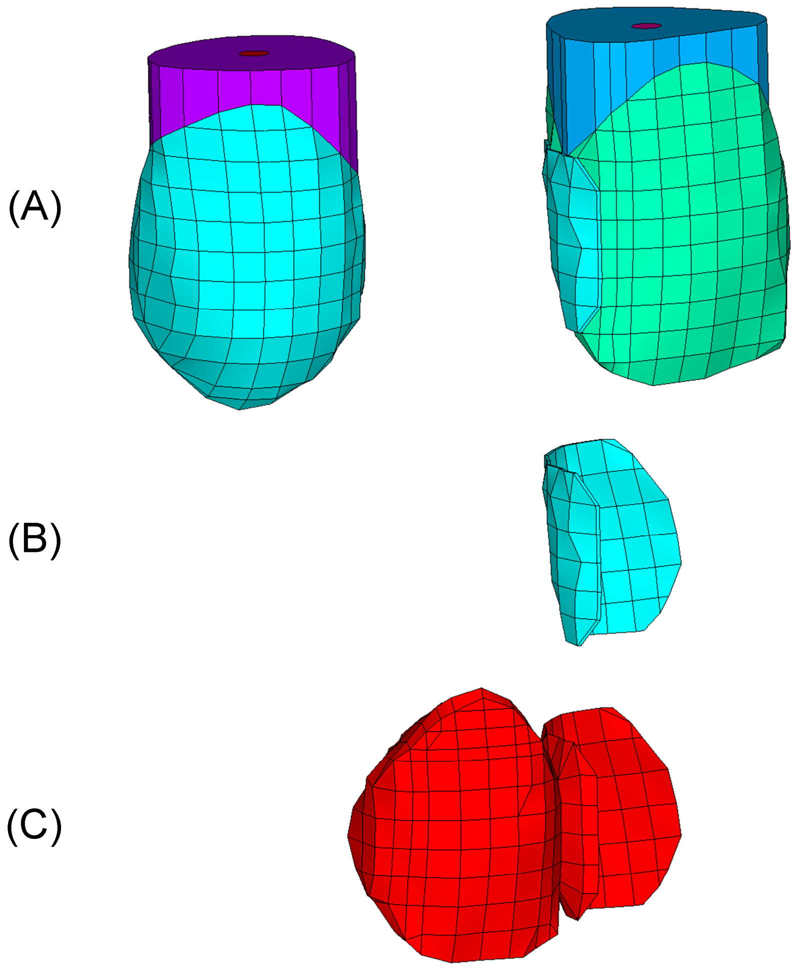

Figure 4.1 3D FE model of a cantilever two-unit RB-FDP: (A) abutment and adjacent tooth, (B) cement layer, (C) RB-FDP.

Materials properties are deviated from clinically used materials (reference brand between parentheses): hybrid particulate filler composite (PFC) for laboratory use (Estenia C&B; Kuraray medical Inc., Tokyo, Japan), hybrid PFC for chairside use (Filtek Z250; 3M ESPE, MN, USA), unidirectional FRC for laboratory use (Estenia C&B EG fiber; Kuraray medical Inc., Tokyo, Japan), unidirectional fibre-reinforced composite for direct and chairside use (everStick C&B; StickTech Ltd., Turku, Finland), Au-Pd alloy (Olympia; J.F. Jelenko, Armork, NY, USA), lithium disilicate glass-ceramic (IPS Empress 2; Ivoclar-Vivadent, Schaan, Liechtenstein), zirconia (InCeram Zirconia; Vita, Bad Säckingen, Germany), feldspathic porcelain (Creation; Klema, Meiningen, Austria), resin-based luting cement (Variolink 2; Ivoclar-Vivadent, Schaan, Liechtenstein), enamel, dentin and pulp. The material properties, mostly obtained from existing literature, are summarised in Table 4.1. The materials were assumed to be isotropic, homogeneous, and linear-elastic, expect for the FRC. The mechanical behaviour of a unidirectional continuous FRC, influenced by their anisotropic (orthotropic) properties, can be described by 3 young's moduli, 3 Poisson's ratios and 3 shear moduli [19]. Twenty-node brick element as solid 95 in ANSYS has the anisotropic material option. Anisotropic material directions corresponded to the element coordinate directions. The orientation of the element coordinate system was altered in such a way it matched the fibre direction.

Table 4.1 Elastic properties of the materials used in the FE model.

	E modulus (GPa)	Poisson's ratio	Shear modulus (MPa)	References
Enamel	80.0	0.30	-	[20]
Dentin	17.6	0.25	-	[21]
Pulp	0.002	0.45	-	[22,23]
Resin luting cement	8.3	0.24	-	[24]
Chairside PFC	11.5	0.31	-	[25,26]
Laboratory PFC	22.0	0.27	-	[19]
Chairside FRC				
longitudinal (X)	46.0	0.39	16.5	a
transverse (Y,Z)	7.0	0.29	2.7	
Laboratory FRC				
longitudinal (X)	39.0	0.35	14.0	[19]
transverse (X,Y)	12.0	0.11	5.4	
Lithium disilicate glass-ceramic	96.0	0.25	-	[24]
Zirconia	205	0.22	-	[24]
Au-Pd alloy	103	0.33	-	[27,28]

a Data obtained by StickTech Ltd. (Turku, Finland)

Five different two-unit cantilever RB-FDP models of various framework materials were generated:

1) FRC-Z250: a FRC-FDP made of a continuous unidirectional E-glass FRC framework (Figure 4.2) veneered with hybrid PFC for direct and chairside use;
2) FRC-ES: a FRC-FDP made of a continuous unidirectional E-glass FRC framework veneered with hybrid PFC for laboratory use;
3) M: a metal-ceramic FDP made of type 3 Au-Pd alloy framework veneered with feldspathic porcelain;
4) GC: an all-ceramic FDP made of a lithium disilicate glass-ceramic framework veneered with feldspathic porcelain;
5) ZI: an all-ceramic FDP made of a zirconia framework and veneered with feldspathic porcelain.

A FRC framework was designed with thickness of 0.6 mm and a height of 3.0 mm [29]. The three-dimensional FE model of the FRC framework and its position in relation to the RB-FDP is shown in Figure 4.2.

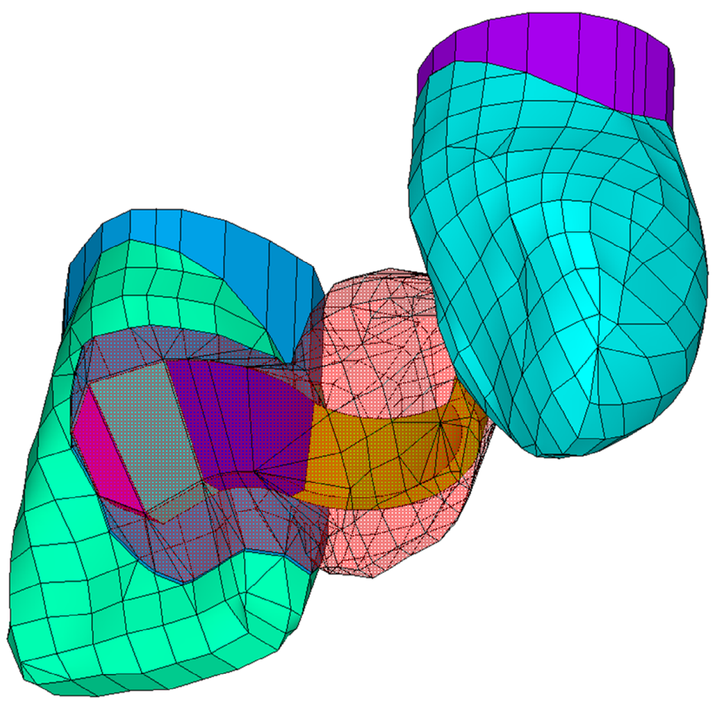

Figure 4.2 3D FE model of a two-unit cantilever FRC RB-FDP: position of the FRC framework in relation to the FDP and the abutment teeth is shown.

Mesh generation, b*oundary conditions, and data processing*

In order to avoid quantitative differences in the stress value in the models, all solid models were derived from a single mapping mesh pattern that generated 103,861 twenty-node brick element (Solid 95 in ANSYS) and 154,784 nodes. The loading and boundary conditions are depicted in Figure 4.3. A stress of 90 MPa was applied at a 45° angle to the incisal edge of the pontic. The final element in all directions of FE model abutment tooth was fixed and distal direction of contact area to canine was fixed. FE analysis was presumed to be linear static. FE model construction and FE analysis were performed on PC workstation (Precision Work Station M90, Dell Inc., Texas, USA) using FE analysis software ANSYS 11. The locations and magnitudes of the principal stress (MPa) and displacement (mm) were identified and used for evaluating the biomechanical behaviour. Maximum principal stress describes the highest in-plane stress and can be regarded to be a tensile stress.

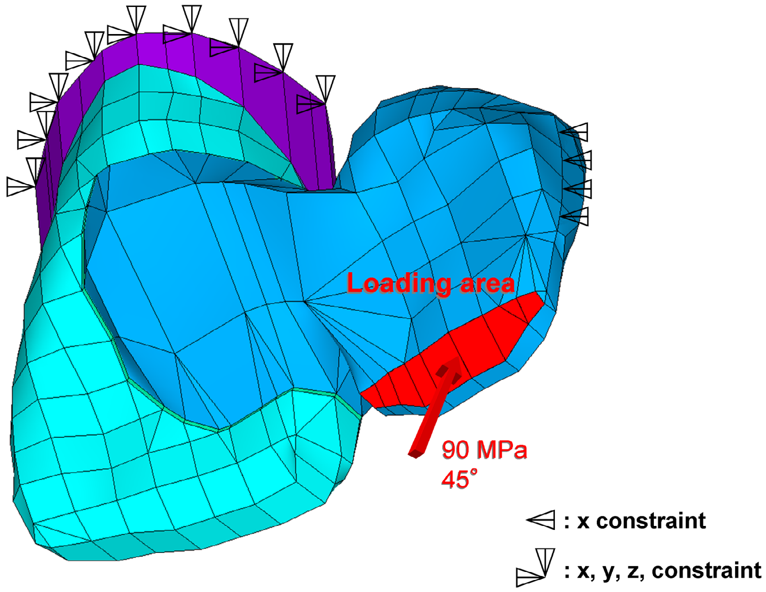

Figure 4.3 Loading and boundary conditions of a 3D FE model representing a two-unit cantilever RB-FDPs.

4.4 Results

Stresses in the FDP

Differences in maximum principal stress were observed (Figure 4.4 and Table 4.2) between the different framework materials and showed a decreasing order: ZI (239.6 MPa) > M (197.1 MPa) > GC (178.4 MPa) > FRC-ES (177.1 MPa) > FRC-Z250 (156.9 MPa). Maximum principal stress concentrations were located in the connector area, more precisely at the occlusal embrasure, for all framework materials. However, additional stress concentrations were observed at the contact area with the adjacent tooth for all framework materials and at the mesio-cervical edge of the retainer for GC (20-30 MPa), M (30-40 MPa) and ZI (50-70 MPa). The principal stresses at the contact area with the adjacent tooth were lower for FRC-ES and FRC-Z250 (30-40 MPa) in comparison to GC (50-70 MPa), M and ZI (>70 MPa).

Stresses at the cement-retainer interface

Differences in maximal principal stress were also observed (Figure 4.5 and Table 4.2) at the cement-retainer interface between all framework materials and showed a decreasing order: ZI (60.8 MPa) > M (36.1 MPa) > GC (32.7 MPa) > FRC-ES (23.9 MPa) > FRC-Z250 (17.5 MPa). Their location differed between all the framework materials. Stress concentrations were observed in the upper part of the proximal area for FRC-Z250 and FRC-ES, while they were located in a semi-circular way around the connector and at the cervical edge of the retainer for M, GC and ZI.

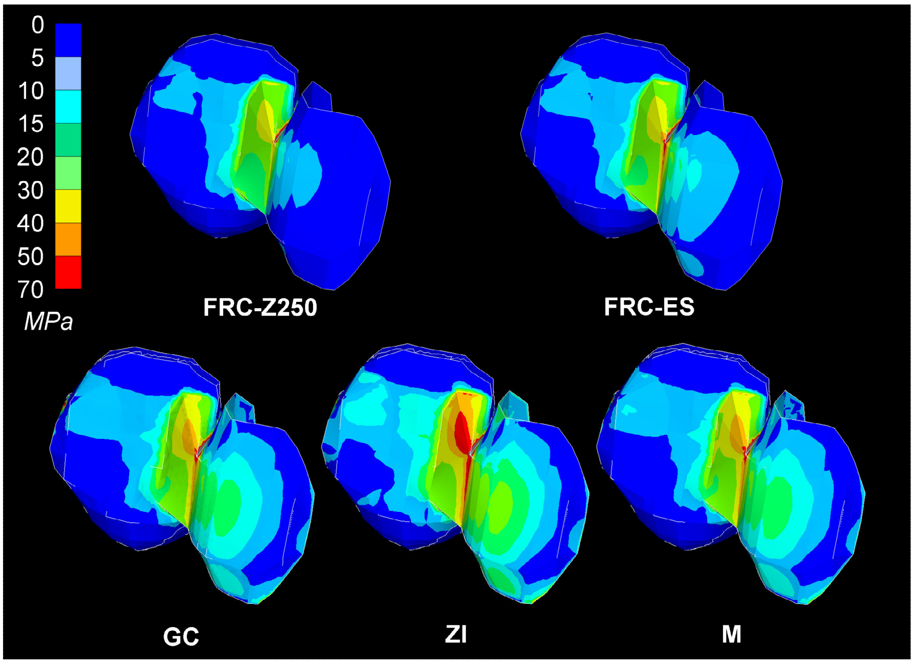

Figure 4.4 Principal stress distribution within two-unit cantilever RB-FDPs of various framework materials.

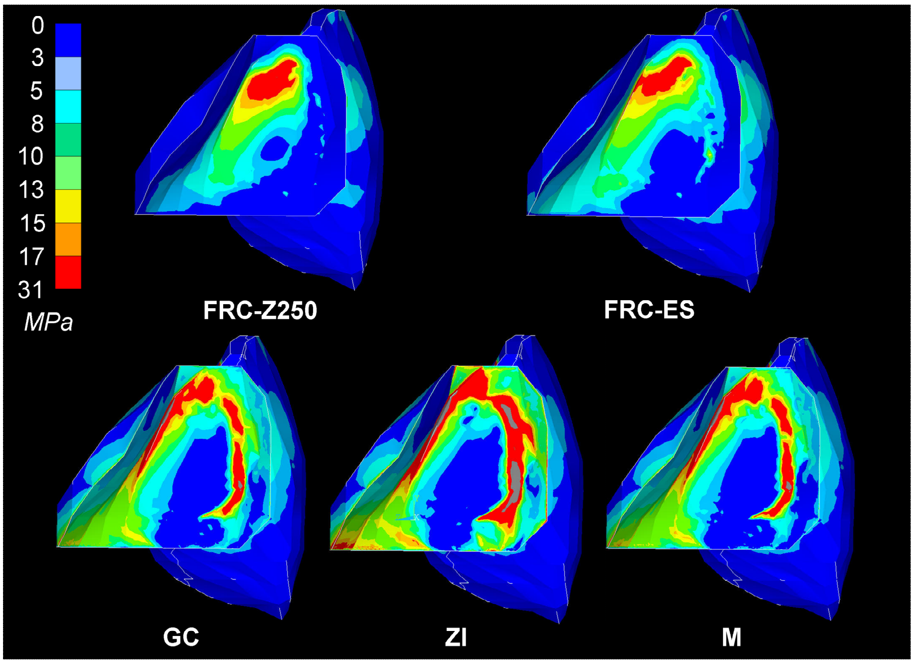

Figure 4.5 Principal stress distribution at the cement-retainer interface for two-unit cantilever RB-FDPs of various framework materials.

Stresses in the cement layer.

FEA revealed (Figure 4.6 and Table 4.2) only slight differences in maximal principal stress between all framework materials and showed a decreasing order: FRC-Z250 (31.3 MPa) > ZI (27.5 MPa) > FRC-ES (27.3 MPa) > M (24.5 MPa) > GC (23.7 MPa). However, they were located in a different area of the cement layer. Highest stress concentrations were located in the upper part of the proximal area for FRC-Z250 and FRC-ES, while they were located at the cervical margin for M, GC and ZI.

Stresses in the abutment tooth

At the abutment tooth only slight differences in maximal principal stress were observed (Figure 4.7 and Table 4.2) between the different framework materials. Highest value was 34.9 MPa for FRC-Z250 and the lowest value was 30.9 MPa for FRC-ES. Once again, their location showed some differences. Highest maximal principal stress concentrations for FRC-Z250 and FRC-ES were observed at the upper middle part of the proximal area and were surrounded by a large area of stress concentration (17-31 MPa) which extended into the palato-cervical area. Highest maximal principal stress concentrations, on the other hand, for M, GC and ZI were located in a small region of the palato-cervical area of the abutment tooth.

Displacement

Differences in maximum displacement were observed in the pontic part of the RB-FDP between the different materials (Table 4.2). Higher displacement of the RB-FDP was encountered with FRC-Z250 (0.048 mm) and FRC-ES (0.035 mm) then with M (0.019 mm), GC (0.019 mm), and ZI (0.017 mm). Although, the maximum displacement of the retainer, cement layer, and abutment tooth revealed the same trend as those for RB-FDPs, a difference of 0.001 mm between highest (0.010 mm) and lowest (0.009 mm) value could not be regarded as clinically relevant. For that reason, maximum displacements of the retainer, cement layer and abutment tooth were regarded similar for all different materials.

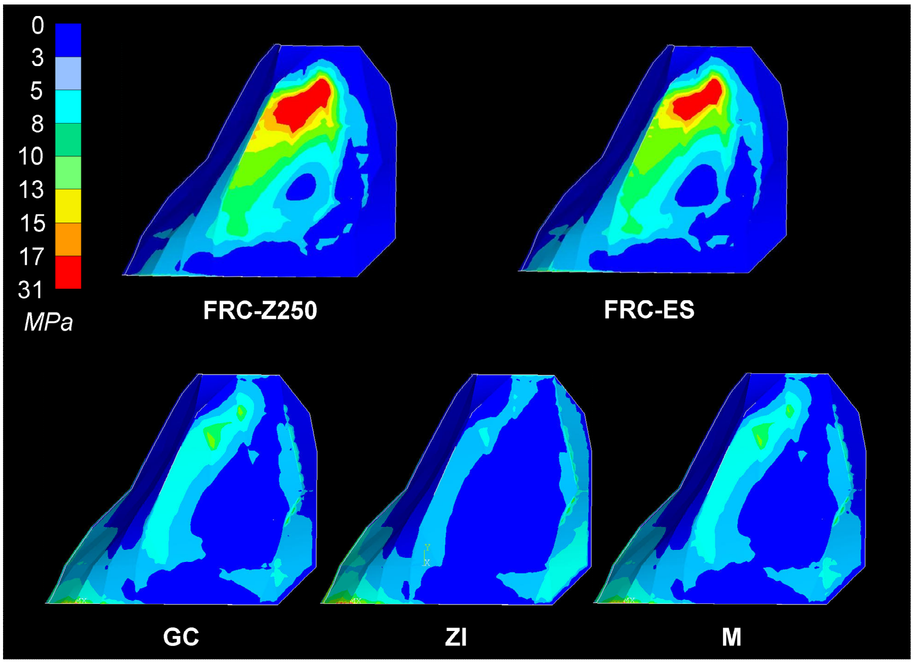

Figure 4.6 Principal stress distribution within the cement layer for two-unit cantilever RB-FDPs of various framework materials.

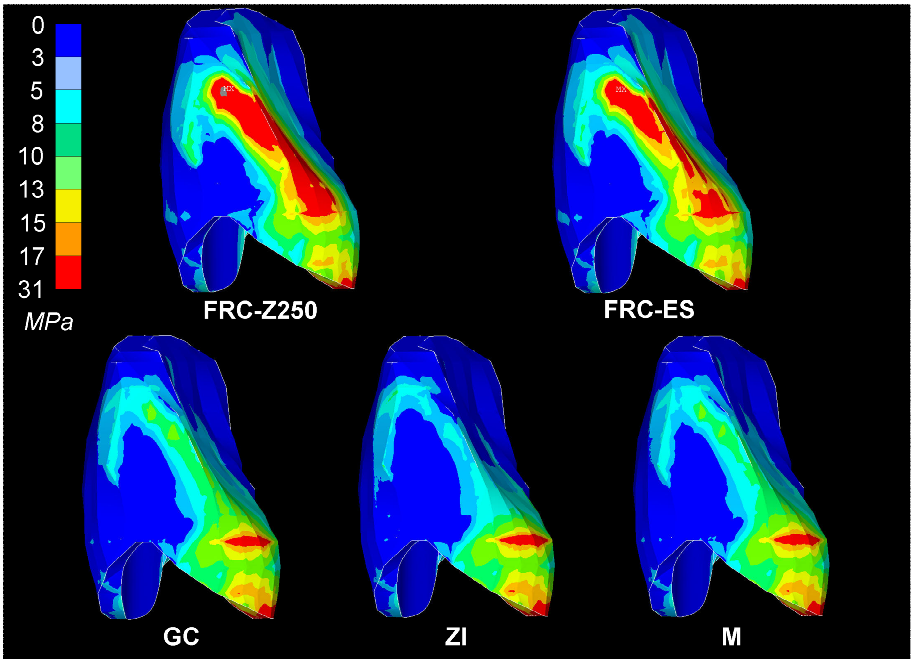

Figure 4.7 Principal stress distribution at the abutment tooth for two-unit cantilever RB-FDPs of various framework materials.

4.5 Discussion

A static fracture strength test, during which a FDP is vertically loaded till failure, is the most common way to evaluate the mechanical behaviour of FDPs in laboratory conditions. The drawbacks of this approach are reckoned by researchers familiar with it. One of these drawbacks is the difficulty to fabricate uniform FDPs in terms of shape and dimensions. Although, FEA can be regarded as a relative easy and cost-effective way to evaluate the mechanical behaviour of complex structures, some limitations of our approach should be acknowledged. Some of these limitations can be drawn back to the simplifications made to the finite element models, eg, tooth model without roots, periodontal ligament [30] and, bone, and the assumptions made related to the material properties [31]. The latter illustrated by the fact that all materials, except FRC, were assumed to be isotropic, homogenous and linear elastic, despite the anisotropic nature of tooth tissue like dentine [32]. Therefore, one should be aware of the fact that the reported values regarding principal stress and displacement can not be regarded as absolute values, which was not the aim of this study. The main purpose of this study was to compare the biomechanical behaviour of anterior two-unit cantilever RB-FDP made of different framework materials. Nevertheless, the ideal approach is to use the results from both FEA and mechanical testing simultaneously, which may be able to provide more reliable and validated data than either method alone [33]. So mechanical testing on two-unit cantilever RB-FDPs in the same condition as this study could be a valuable asset.

In the present study, the FE model was loaded by applying a stress of 90 MPa in a 45° angle to the incisal edge of the pontic tooth. An applied stress of 90 MPa to a 5.5 mm^2 incisal area corresponds to a load of 495 N. The applied load is significantly higher than previously reported maximum anterior mastication loads of 108-382 N [34,35] and therefore can be regarded as the worst case scenario. In clinical circumstances, an anterior occlusal contact more closely resembles an area than a point, for that reason it was chosen to apply the load to a loading area.

Roots, periodontal ligament and bone, which are responsible for physiologic tooth mobility, were not included in the FE model. Under clinical conditions, a part of the loading is transferred via the roots and the periodontal ligament into the bone. The lack of physiologic tooth mobility in the present FE model negatively influences the outcome of the FEA, in such a way the principal stress values are overestimated. The effect of tooth mobility was illustrated by Rosentritt *et al.* [36], who found higher fracture strengths for anterior cantilever RB-FDPs when luted to abutment teeth with high mobility [36]. Clinically, the rationale to use a cantilever design instead of fixed-fixed design is related to the teeth mobility. The risk for debonding of three-unit fixed-fixed RB-FDP from one end is relatively high, when teeth with increased mobility are involved abutment. A debonded retainer may result in secondary caries which is not diagnosed in time.

The present FEA revealed differences in biomechanical behaviour, more precisely stress distribution and displacement, between RB-FDPs made of different framework materials (Table 4.2).

Although the location of the maximum principal stresses and displacement, observed at the FDP level, was identical for all framework materials, the values differed. The differences in displacement and principal stress can be explained by the differences in elastic modulus (stiffness) between the framework materials. RB-FDPs made of materials with a higher stiffness suffered less displacement, but higher principal stress than those made of less stiff materials, which can be illustrated by comparison of zirconia and chairside FRC. Zirconia has a elastic modulus of 205 GPa and showed 0.017 mm displacement with 239.6 MPa maximum principal stress in comparison to the 0.048 mm and the 156.9 MPa by the chairside FRC with an elastic modulus between 11 GPa (chairside hybrid composite) and 46 GPa (FRC). The highest maximum principal stress was located at the occlusal embrasure of the connector. It has to be noticed that the connector in our FE model was designed with a sharp embrasure and that stresses at the occlusal embrasure of the connector can be significantly decreased by changing the connector design [37] and the radius of curvature of the connector strongly affects the fracture resistance of a FDP [37,38]. Recently, Plengsombut *et al*. confirmed this finding by revealing a significant lower fracture strength for specimens with a round connector in comparison to those with a sharp connector [39].

A comparable situation with regard to stress values was found at the level of cement-retainer interface. Far more interesting were the differences in location between FRC on one hand and M, GC and ZI on the other hand (Figure 4.4). A possible explanation is the difference in design between both groups of FDPs. In a FRC-FDP the stiffer fibres transfer the stress from the pontic to the central part of the retainer corresponding to the connector location, in contrast to FDPs (M, GC and ZI) with a uniform framework design were the stress is transferred to an area around the connector and towards the cervical margin of the retainer. Debonding of the FDPs due to premature failure of the adhesive interface between retainer and cement layer, is likely to be caused by such unfavourable stress location in combination with direct

exposure to the oral environment. Especially zirconia, known for it's questionable adhesion to resin luting cements [40,41], will be prone to adhesive failure.
At the level of the cement layer there was a only a slight difference between maximum principal stress values of all framework materials, but as expected the differences in location, as seen at the cement-retainer interface, between FRC on one hand and M, GC and ZI on the other hand (Figure 4.6) became more pronounced at the cement layer. It is interesting to notice that the cement layer, in the case of M,GC and ZI , is able to absorb the stresses in the area surrounding the connector and to dissipate those stresses towards the cervical outline. Such unfavourable stress transfer can result in premature failure of the cement layer.
The difference in maximum principal stress value between different framework materials was even lower at the level of the abutment tooth. However, the location of the stress concentration, as depicted in Figure 4.7, was different. Adhesive failure at the enamel-cement interface is not very likely to occur, as enamel bonding is a reliable procedure with reported values for resin luting cements, like Variolink 2, of 49.3 MPa [42].

Table 4.2 Maximum and minimum principal stress (MPa) and displacement (μm) for two-unit cantilever RB-FDPs of various framework materials.

	FDP			Cement-retainer interface			Cement layer			Abutment tooth		
	max	min	disp	max	min	disp	max	min	disp	max	min	disp
FRC-Z250	156.9	-56.2	48	17.5	-5.3	10	31.3	-7.1	10	34.9	-7.6	10
FRC-ES	177.1	-67.2	35	23.9	-9.7	10	27.3	-7.1	10	30.9	-9.8	10
GC	178.4	-116.3	19	32.7	-42.5	9	23.7	-4.1	9	31.4	-4.8	9
ZI	239.6	-154.3	17	60.8	-75.3	9	27.5	-3.3	9	31.7	-7.2	9
M	197.1	-149.9	19	36.1	-45.8	9	24.5	-3.7	9	31.9	-5.0	9

Based on the results of this study the predominant failure mode of two-unit cantilever RB-FDPs for each framework material might be predicted. Zirconia and metal RB-FDPs are suspected to fail most likely because of debonding. A multitude of clinical research on cantilever metal RB-FDPs corroborates this prediction [7-9], since debonding was reported as the major reason of failure. Metal alloys exhibits plasticity, which can explain this mode of failure. On the other hand, only a limited amount of *in vitro* studies on zirconia RB-FDPs are available. It was shown that minimal invasive cantilever RB-FDPs subjected to fatigue loading, predominantly failed due to debonding [36,43]. However, the same studies showed a decrease in percentage of debonding in favour of retainer fractures, when a more retentive retainer design was used. Although, one should be aware that the high stress concentrations at the mesio-cervical edge of the retainer indicates (Figure 4.5) that retainer fracture is most probably the result of partial debonding. Due to partial debonding more complex torque and bending forces acts on the retainer, which results in retainer fracture.

Glass ceramic and FRC RB-FDPs might be more susceptible for connector fractures. Since no studies on cantilever glass ceramic RB-FDPs are conducted, the only studies available are those on cantilever alumina RB-FDPs [10,44,45]. These cantilever alumina RB-FDPs exhibited a 5-year survival rate of 92.3% [10]. During their study only one cantilever RB-FDP was lost due to fracture of the connector. Koutayas *et al*. reported connector fracture as the predominant fatigue failure of cantilever alumina RB-FDPs [44,45]. Since glass ceramic exhibits flexure strength of 252 MPa [46], which is inferior to the flexure strength of alumina (429 MPa) reported by Tinschert *et al*. [47] and their bond strength to resin luting cements is superior to that of alumina [48], the previous described studies can be regarded as representative for the affirmation of their expected failure mode. Clinical [49] and *in vitro* [50,51] findings on FRC RB-FDPs also confirms this prediction. In comparison to glass ceramic and zirconia, were connector fracture results in an immediate aesthetic problem, this is not the case for FRC. From an aesthetic point of view the fibre reinforcement fulfils a fail-safe situation, because even after connector fracture the fibre reinforcement protects the FDP from complete debonding.

The results of this study on anterior two-unit cantilever RB-FDPs can be compared to those of Shinya *et al*. [17] on anterior three-unit fixed-fixed RB-FDPs. It should be noticed that the FE model and material properties were exactly the same for both studies, but that only FRC-, and metal-based three-unit fixed-fixed RB-FDPs were evaluated by Shinya *et al*. [17]. It is interesting to observe that the difference in principal stresses between various framework materials is higher for three-unit fixed-fixed designs than for two-unit cantilever designs. This suggests that the influence of framework material is less important for two-unit cantilever designs.

Metal-based anterior two-unit cantilever RB-FDPs, proven to be a clinically viable treatment option [4,6-9], can be regarded to be the gold standard for comparison with the other materials. Although acceptable bond strength to resin luting cements can be achieved by glass ceramics, their low strength in combination with the less even stress distribution from loading area towards abutment tooth makes it not to be a suitable material for the fabrication of anterior two-unit cantilever RB-FDPs. FRC-based RB-FDPs seems to be more promising as they exhibits a good bond strength to resin luting cement and more even stress distribution. Nevertheless, they are at the moment only suitable as low cost temporary alternative due to the low strength of the veneering composite. Further improvements can be expected from modified framework designs [52] and improved resin composites [53]. Zirconia, regardless of its high strength, does not seems to be the ideal material for cantilever RB-FDPs, due to the unfavourable stress distribution and low bond strength to resin luting cement leading to premature debonding. Recent improvement of the adhesive performance of zirconia by selective infiltration etching increased the bond strength to Panavia F2.0 up to 49.8 MPa [41]. The achievement of a strong and durable bond with zirconia-based materials, makes it a most promising alternative to metal-based anterior two-unit cantilever RB-FDPs.

4.6 Conclusions

Within the limitations of this study, FEA revealed differences in biomechanical behaviour between RB-FDPs made of different framework materials:

1. The general observation was that a RB-FDP made of FRC provided a more evenly distributed stress pattern from loading area towards abutment tooth.
2. Maximum principal stress was located at the occlusal embrasure of the connector for all framework materials: highest value was found for ZI, while the lowest for FRC-Z250.
3. Advanced stress analyses suggest a possible difference in predominant failure mode: connector fracture for FRC-, and glass ceramic-based RB-FDPs and debonding for metal-, and zirconia-based RB-FDPs.
4. A stress concentration was found at the contact area with the adjacent tooth, indicating that the applied load is partially transferred to the adjacent tooth.

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CHAPTER 5

Retainer design of indirect two-unit cantilever resin-bonded glass fibre-reinforced composite fixed dental prostheses in the premolar region: an in vitro and finite element analysis study

5.1 Abstract

Objectives: The aim of this study was to evaluate in vitro the influence of retainer design on the strength of two-unit cantilever resin-bonded glass fibre-reinforced composite (FRC) fixed dental prostheses (FDP).

Materials and Methods: Four retainer designs were tested: a proximal box, a step-box, a dual wing and a step-box-wing. Of each design on eight human mandibular molars, FRC-FDPs of a premolar size were produced. The FRC framework was made of resin impregnated unidirectional glass fibres (Estenia C&B EG Fiber, Kuraray) and veneered with hybrid resin composite (Estenia C&B, Kuraray). Panavia F2.0 (Kuraray) was used as resin luting cement. FRC-FDPs were loaded to failure in a universal testing machine. One-way ANOVA and Tukey's post hoc test were used to evaluate the data. The four designs were analysed with finite element analysis (FEA) to reveal the stress distribution within the tooth/restoration complex.

Results: Significant lower fracture strengths were observed with inlay-retained FDPs (proximal box: 300 ± 65 N; step-box: 309 ± 37 N) compared to wing-retained FDPs ($p<0.05$) (step-box-wing: 662 ± 99 N; dual wing: 697 ± 67 N). Proximal-box, step-box and step-box-wing-retained FDPs mainly failed with catastrophic cusp fracture (proximal box 100%, step-box 100%, and step-box-wing 75%), while dual wing-retained FDPs mainly failed at the adhesive interface and/or due to pontic failure (75%). FEA showed more favourable stress distributions within the tooth/restoration complex for dual wing retainers.

Conclusions: It was concluded that a dual-wing retainer is the optimal design for replacement of a single premolar by means of a two-unit cantilever FRC-FDPs.

5.2 Introduction

Single-tooth replacement in the anterior and premolar region is more often required to improve aesthetics than for functional reasons. Contemporary dentistry offers a broad range of treatment modalities for single tooth replacement, e.g. autogenous tooth transplantation, removable dental prostheses (RDPs), fixed dental prostheses (FDPs), and implants. Although autogenous tooth transplantation and RDPs are viable treatment options from the point of view of preserving tooth tissue and reduction of cost, their indication and use are limited [1,2]. Instead, three-unit fixed dental prostheses and implant-retained crowns are acknowledged as the treatment of choice [3,4]. In cases with limited bone height and/or width and extensively restored adjacent teeth, a FDP is preferred, while implant-retained crowns are chosen when neighbouring teeth are free of restorations and/or caries. However, not all single-tooth gaps can be restored by means of conventional FDPs or implant-retained crowns.

In cases involving patients with diastema less than 7 mm and caries-free adjacent teeth, or those with reduced financial resources, resin-bonded fixed dental prostheses (RB-FDP) have proved to be a reliable alternative [5]. Nevertheless, metal ceramic RB-FDPs have some drawbacks, such as the greyish appearance of abutment teeth caused by shine-through of metal retainers. Another common problem with RB-FDPs is early loss of retention caused by the number of abutments and a lack of retentive and resistant preparation [5-7].

Clinical research has shown that in order to improve retention and resistance and the subsequent longevity of RB-FDPs, the abutment teeth need more extensive preparation; this should include not only complete palatal or lingual coverage with 180-degree wraparound, but also chamfer, occlusal or cingulum rests, and proximal guide planes and grooves [5,6,8-10].

In particular, it is often the case that only one of the retainers debonds [11]. After removal of the debonded retainer, many of these partially debonded bridges have successfully converted into a cantilever design [12]. Dynamic tooth contacts are believed to induce twisting and shear forces which cause retainers in fixed-fixed RB-FDPs to be dislodged; this is referred to as biting the tooth out of the retainer [6,7,9,13,14]. The free-standing nature of two-unit cantilever RB-FDPs is thought to reduce or even eliminate these adverse stresses on the adhesive interface during function [6,9,13]. Clinical research has demonstrated that two-unit cantilever RB-FDPs performed as well as or even better as their three-unit fixed-fixed counterparts [7,12,14-16].

Over the last few years, fibre-reinforced composites (FRC) have become more popular [17]. The introduction and subsequent development of adhesive dentistry established the paradigm shift from G.V. Black's "extension for prevention"[18] to minimal invasive dentistry [19,20]. The interest in metal-free FDPs was stimulated particularly by the less acceptable aesthetics of metal ceramic FDPs, and by growing awareness in the dental profession of allergic reactions to dental alloys [21]. This continuous search for less invasive and metal-free treatments focused attention on fibre-reinforced composite fixed dental prostheses (FRC-FDPs), whose current popularity can be attributed to the fact they can be fabricated not only in the dental laboratory, but also at the chairside by the dentist. Clinical trials with evaluation periods of up to five years have demonstrated that FRC-FDPs are indeed a suitable treatment option [22-25]: even a longevity of at least ten years now seems reasonable [26].

Dentistry has now entered an era in which preservation of tooth tissue and aesthetics are of utmost importance when restoring the dentition. A two-unit cantilever resin-bonded FRC-FDP is one such conservative and aesthetic alternative.

To our knowledge only three publications have reported on this treatment modality for anterior single-tooth replacement [27-29]. A clinical report by Culy *et al.* [29] concluded after only 10 months of observation that direct cantilever resin-bonded FRC-FDPs could be a viable option for replacing anterior teeth. Li *et al.* [27,28] determined failure load, deflection and failure location, and identified the role of the fibres and the adjacent teeth in an *in vitro* study and in a finite element analysis (FEA) study.

Not only two-unit cantilever metal ceramic RB-FDPs are proven to be a predictable and successful prosthetic reconstruction in the anterior and posterior region in the short to medium term [14,16], but also two-unit cantilever resin-bonded FRC-FDPs could be a viable anterior single-tooth replacement [29].

The aim of the present study was to investigate *in vitro* the influence of retainer design on the strength and stress distribution in the tooth/restoration complex of indirect two-unit cantilever resin-bonded glass fibre-reinforced composite fixed dental prostheses in the premolar region. Four different retainer designs were compared. A static fracture strength test was conducted to evaluate the strength of these restorations. Stress distribution in the tooth/restoration complex was analysed by means of 3D FEA.

5.3 Materials and Methods

Table 5.1 Materials used for static fracture strength test of two-unit cantilever resin-bonded FRC-FDPs.

Brand	Composition	Manufacturer	Lot number
Estenia C&B EG Fiber	UTMA, silanised E-glass fibres, ultra fine silica filler	Kuraray medical Inc, Okayama, Japan	0003AB
Estenia C&B Dentine A2	UTMA, Bis-GMA, TEGDMA, glass ceramic, Al_2O_3	Kuraray medical Inc, Okayama, Japan	00219A
Panavia F2.0 ED II Primer	EDII primer and luting resin Primer A: HEMA, MDP, 5-NMSA, water, accelerator Primer B: 5-NMSA, accelerator, water, sodium benzene sulphinate	Kuraray medical Inc, Okayama, Japan	41170
Luting resin	Base paste: hydrophobic aromatic (and aliphatic) dimethacrylate, hydrophilic dimethacrylate, sodium aromatic sulfinate, N,N-diethanol-p-toluidine, functionalized sodium fluoride, silanized barium glass Catalyst paste: MDP, hydrophobic aromatic (and aliphatic) dimethacrylate, hydrophilic dimethacrylate, silanized silica, photoinitator, dibenzoyl peroxide		
Clearfil Porcelain Bond Activator	Hydrophobic dimethacrylate, MPTS, Bis-PMA	Kuraray medical Inc, Okayama, Japan	00158B
Clearfil SE Bond Primer	MDP, HEMA, hydrophilic dimethacrylate, dl-camphorquinone, water	Kuraray medical Inc, Okayama, Japan	00407A

UTMA urethane teramethacrylate; Bis-GMA bisphenol-A-glycidyl dimethacrylate; TEGDMA triethylenglycol dimethacrylate; MDP 10-methacryloyloxydecyl dihydrogen phosphate; HEMA 2-hydroxyethyl methacrylate; 5-NMSA N-methacrylyloyl 5-aminosalicylic acid; MPTS 3-methhacryloxypropyl trimethoxy silane; Bis-PMA bisphenol-A-polyethoxy dimethacrylate.

Molar-borne two-unit cantilever resin-bonded FRC-FDPs were constructed according to various retainer designs. A static fracture strength test was conducted to evaluate the strength of these restorations. Stress distribution in the tooth\restoration complex was analysed by means of 3D FEA. A recently introduced all-resin restorative system for the fabrication of laboratory-made crown and bridgework was used for this experiment, the restorative system was composed of a new generation hybrid resin-based composite (Estenia C&B), a proprietary glass fibre-reinforcement (Estenia C&B EG Fiber) and a dual-cured resin luting cement (Panavia F2.0). EG

Fiber contains 48 wt% silanised E-glass fibres of 11 μm in diameter impregnated into an urethane tetramethacrylate-based resin [30,31]. The composition of the materials used in this study is summarised in Table 5.1.

Fracture Strength

Thirty-two freshly extracted human mandibular molars without caries or restorations were selected and stored in tap water at 5°C prior to use. Each tooth was positioned into a copper pipe and embedded in poly(methyl methacrylate) resin (Vertex self curing, Vertex-Dental BV, Zeist, the Netherlands) within 2 mm from the cemento-enamel junction. The specimen were randomly divided in four groups (n = 8) and stored in tap water at 5°C until use.

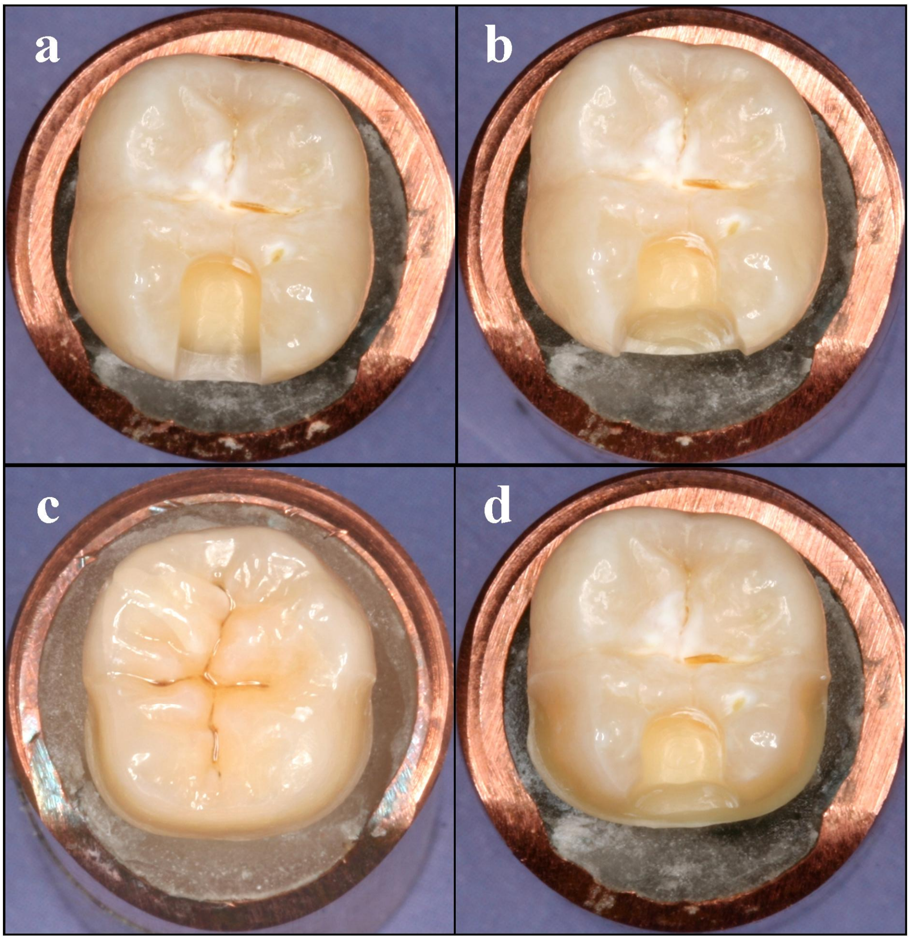

Figure 5.1 Different types of retainer preparation: (a) proximal box preparation (2 mm high, 2 mm wide, 4 mm deep), (b) step-box preparation (step: 2 mm high, 2mm wide, 4 mm deep; box: 3.5 mm high, 3.5 mm wide, 1.5 mm deep), (c) dual wing preparation (4 mm), and (d) step-box-wing preparation.

Four different retainer designs were tested (Figure 5.1): a proximal box preparation (2 mm high, 2 mm wide, 4 mm deep), a step-box preparation (step: 2 mm high, 2mm wide, 4 mm deep; box: 3.5 mm high, 3.5 mm wide, 1.5 mm deep), a dual-wing preparation, which consisted of a vestibular and lingual adhesive wing (4 mm long), and a step-box-wing preparation which is the combination of a step-box and a dual wing. Proximal-box-retained and step-box-retained FDPs will be referred as inlay-retained FDPs, while step-box-wing and dual-wing-retained FDPs as wing-retained FDPs.

All preparations were made by a single operator using conventional diamond burs (preparation set 4278 and 4384A, Komet, Lemgo, Germany) in a water-cooled, high speed contra-angle handpiece (Kavo Dental, Biberach/Riss, Germany). The dimensions of the preparation were measured with a digital calliper (digimatic, Mitutoyo, Kawasaki, Japan) and standardised by minor adjustments.

Two-unit cantilever resin-bonded FRC-FDPs were fabricated according to the indirect technique. The FRC framework was made of resin pre-impregnated unidirectional E-glass fibres (Estenia C&B EG Fiber); one bundle of EG Fiber consisting of about 15,000 glass fibres. While the framework of inlay-retained FDPs was reinforced with one bundle of FRC, two bundles were used in the framework of wing-retained FDPs. Fibre-reinforcement was placed in the area of the FDP were tensile stresses were expected to occur; for cantilever restorations these area is situated near the occlusal surface. The fibre location throughout the FDPs is shown in Figure 5.2. The FRC framework was light polymerised for 10 s with a handheld polymerisation unit (Astralis 10, Ivoclar-Vivadent, Schaan, Liechtenstein) with a power output of 1000 $mW\cdot cm^{-2}$ (Curing Radiometer model 100, Demetron research corporation, Danbury, USA).

The retainer and the premolar pontic were veneered in increments with hybrid particulate filler composite (PFC) for indirect use (Estenia C&B, shade dentine A2). A poly(vinyl siloxane) template was used to standardise the dimensions of each FDP (pontic: 8 mm high, 8.5 mm wide in buccal-lingual direction, and 7 mm wide in mesial-distal direction). The connector size differed according to the number of FRC-bundles: 5 mm wide and 5 mm high for the inlay-retained FDPs and 6.5 mm wide and 5.5 mm high for the wing-retained FDPs. Each increment was light polymerised for 10 s. The completed FDP was post polymerised by light and heat in a light furnace (Lumamat 100, Program 1, Ivoclar-Vivadent, Schaan, Liechtenstein) for 25 min. The FRC-FDPs were luted with a MDP-monomer containing resin luting cement (Panavia F 2.0, shade TC) according manufacturer's instructions.

After one week water storage at 37ºC, the specimens were loaded to failure in a universal testing machine (Instron 6022, Instron Limited, Wycombe, UK). The load was applied to the central fossa of the premolar pontic by a steel contact ball of 6 mm in diameter at a crosshead speed of 1 $mm \cdot min^{-1}$.

All fractured specimens were visually examined and their mode of failure was recorded. Adhesive failures were further examined under a light microscope (4x magnification).

Finite Element Analysis

Three-dimensional simplified finite element models were created of a two-unit mesial cantilever on a mandibular first molar. Both the molar and the pontic were 8 mm high, 10.5 mm wide in the buccal-lingual direction, and the molar being 11 mm and the pontic being 7 mm wide in the mesial-distal direction. The root of the molar was 10 mm in length. The retainer designs were the same as those used for the fracture strength test. The finite element modelling was carried out with FEMAP software (FEMAP 8.10, ESP, Maryland Height, MO, USA), while the analysis was carried out with CAEFEM 7.3 (CAC, West Hills, CA, USA). The models were composed of 57,000-66,000 parabolic tetrahedron solid elements. The material properties are summarised in Table 5.2, with the exception of the FRC, these properties were assumed to be isotropic, homogenous and linear-elastic. Material properties data for Estenia C&B and Estenia C&B EG Fiber were provided by the manufacturer; the data for dentine were obtained from existing literature [32]. The nodes at the bottom of the root were fixed (no translation or rotation in any direction).

A load of 300 N was applied at the centre of the pontic for the proximal-box-retained FDPs and the step-box-retained FDPs. For dual-wing-retained and step-box-wing-retained FDPs a load of 650 N was applied. Two stresses were calculated to establish the peel-off stress on the major attachment surfaces: the Solid Major Principle stress and the Solid S_x stress: the peel-off stress is defined as the tensile stress perpendicular to the bonding surface.

Table 5.2 Material properties used in the 3D FEA model

Material	Product	Elastic modulus (GPa)	Shear modulus (GPa)	Poison's ratio
Dentine		18		0.31
Composite	Estenia C&B	22		0.27
Fibre-reinforced composite	Estenia C&B EG fiber	39 12 12	14 5.4 5.4	0.35 0.11 0.11

Statistical Analyses

Statistical analysis was performed with the statistical software SPSS for windows 12.0.1 (SPSS Inc. Chicago, IL, USA). Mean and standard deviations of fracture strength for each group were calculated. One-way analysis of variance (ANOVA) followed by Tukey's post hoc test was performed to determine the effect of retainer design on the fracture strengths observed. P-values of less than 0.05 were considered to be statistically significant.

5.4 Results

Fracture Strength

One-way ANOVA ($F = 75.32$; $p < 0.001$; power = 1.0) revealed that the retainer design had a statistical significant effect on the static fracture strength of two-unit cantilever resin-bonded FRC-FDPs. However, Tukey's multiple comparison test ($p < 0.001$) showed only significant differences between inlay-retained designs and wing-retained designs (Figure 2). The proximal-box-retained design yielded the lowest mean fracture strength, which was not significant different ($p = 0.993$) from the step-box-retained design, respectively 300 (65) N and 309 (37) N. Significantly higher mean fracture strengths were obtained with wing-retained FDPs ($p < 0.001$). The dual-wing-retained design showed slightly, but not significantly ($p = 0.746$), higher fracture strength values than the step-box-wing-retained design, respectively 697 (67) N and 662 (99) N. The results of the fracture strength test are graphically presented in Figure 5.3.

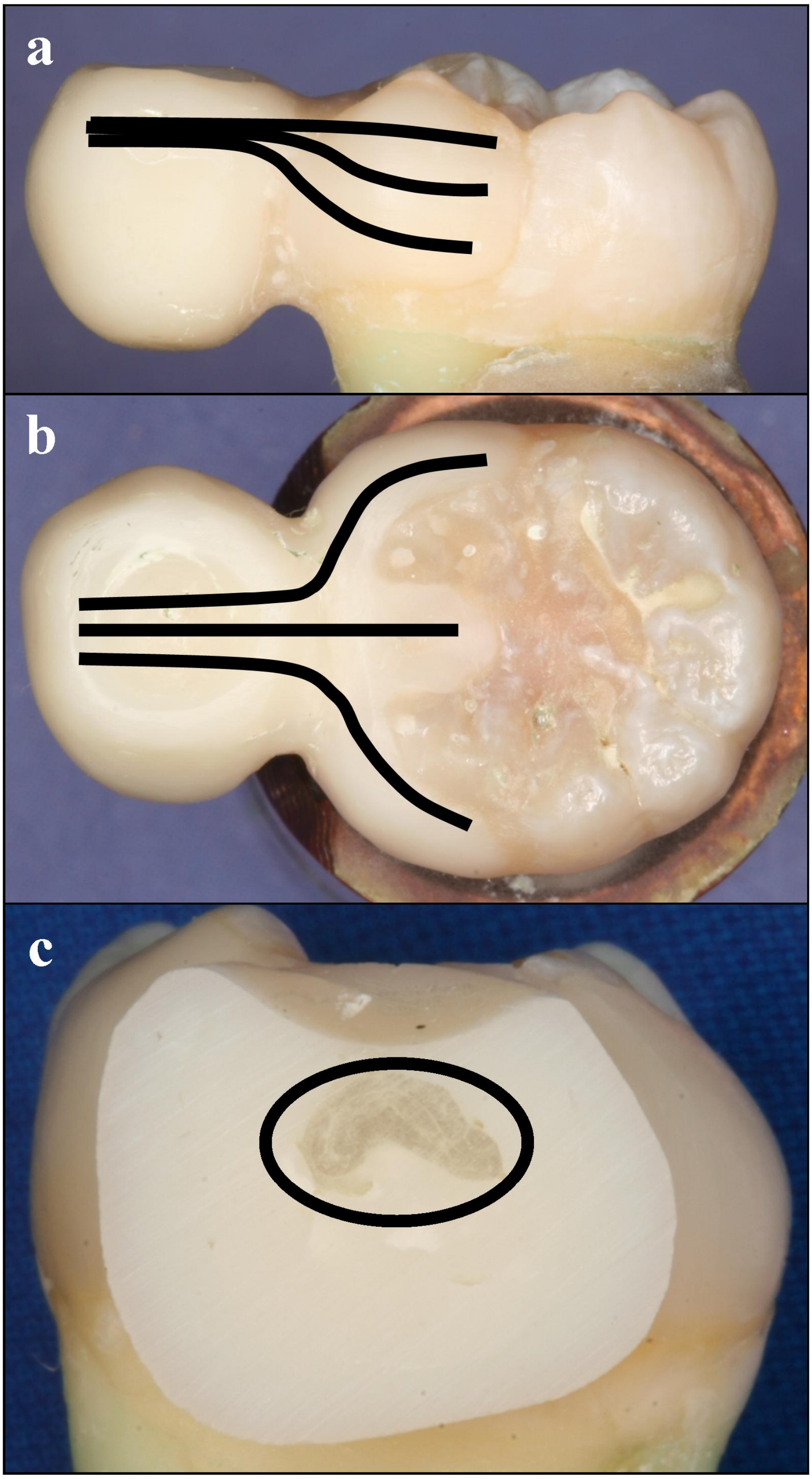

Figure 5.2 Fibre location (black lines) throughout a two-unit cantilever resin-bonded FRC-FDP: (a) longitudinal view, (b) occlusal view, and (c) cross-sectional view through the pontic.

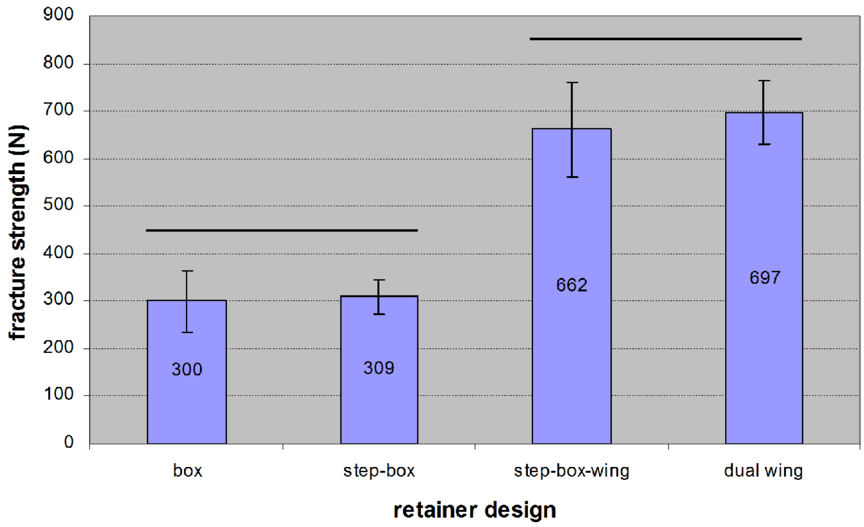

Figure 5.3 Bar diagram of static fracture strength (N) of two-unit cantilever resin-bonded FRC-FDPs with the mean (number) and standard deviations (error bars) for the four different retainer designs. There is no statistical significant difference between groups beneath the horizontal line.

The failure modes of the FRC-FDPs and their distribution are given in Table 5.3. Four modes of failure were observed: tooth fracture, FDP fracture, adhesive failure, and a combination of adhesive failure and FDP fracture. The predominant modes of failure of two-unit cantilever resin-bonded FRC-FDPs are shown in Figure 5.4. Failure mode analysis showed that inlay-retained FDPs all failed because of tooth fracture. On the other hand, hundred percent of the step-box-wing-retained FDPs failed because of catastrophic cusp fracture. Only fifty percent of the specimen in the proximal-box-retained group and the step-box-retained group, which failed because of tooth fracture, really suffered from catastrophic cusp fracture, while in the step-box-wing-retained group all these specimens failed because of cusp fracture. Seventy-five percent of the dual-wing-retained FDPs failed at the adhesive interface and/or due to pontic failure. Closer inspection of the adhesively fractured FDPs revealed that these specimens failed not only adhesively between luting agent and enamel, but also at the luting-Estenia interface.

Table 5.3 Modes of failure for two-unit cantilever resin-bonded FRC-FDPs.

Retainer design	Tooth fracture (%)	FDP fracture (%)	Adhesive failure (%)	Combination adhesive failure and FDP fracture (%)
Proximal box	8 (100)	0 (0)	0 (0)	0 (0)
Step-box	8 (100)	0 (0)	0 (0)	0 (0)
Step-box-wing	6 (75)	1 (12.5)	1 (12.5)	0 (0)
Dual wing	2 (25)	1 (12.5)	3 (37.5)	2 (25)

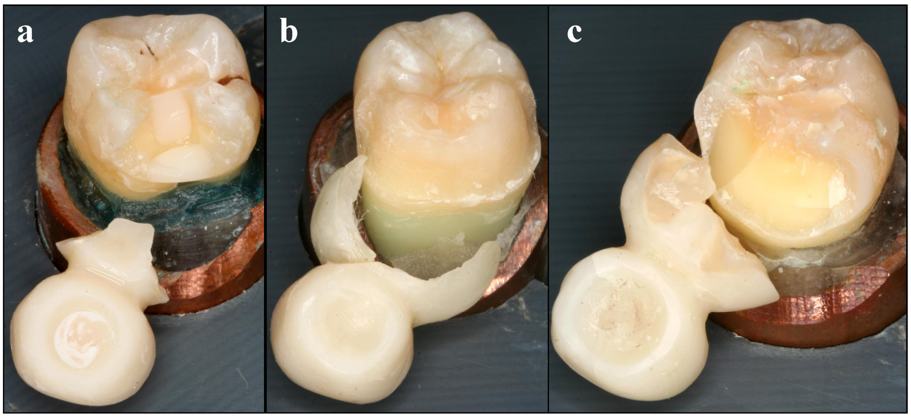

Figure 5.4 Predominant modes of failure of two-unit cantilever resin-bonded FRC-FDPs: (a) tooth fracture for inlay-retained design (proximal box and step-box), (b) adhesive failure for dual-wing-retained design, and (c) catastrophic cusp fracture for step-box-wing-retained design.

Finite Element Analyses

The results of the FEA with the 300 N load on the inlay-retained FDPs, and the 650 N load on the wing-retained FDPs are presented in Table 5.4 showing the maximum Solid Major Principle Stress in the tooth and the maximum Solid S_x (peel-off stress) on the proximal contact area. Stress distribution within the tooth and the FRC framework for the four retainer designs are shown in Figure 5.5. For the inlay-retained FDPs, highest tensile stresses and peel-off stresses were encountered at the proximal surface on the left-hand and the right-hand side of the box preparation. With step-box-wing-retained FDPs the highest tensile stresses presented at the central groove of the occlusal surface, while the highest peel-off stresses were found at the left and right proximal surface of the box preparation. In case of wing-retained FDPs, the highest tensile as well as peel-off stresses are seen in the same area, namely in the occlusal part of the proximal surface.

Table 5.4 Maximum stresses with the different retainer designs

	Max. Solid Major Principle Stress (MPa)	Max. Solid Sx Stress (MPa)
Proximalbox	66.4	40.5
Step-box	70.0	46.3
Dual wing	52.3	48.8
Step-box- wing	56.8	44.3

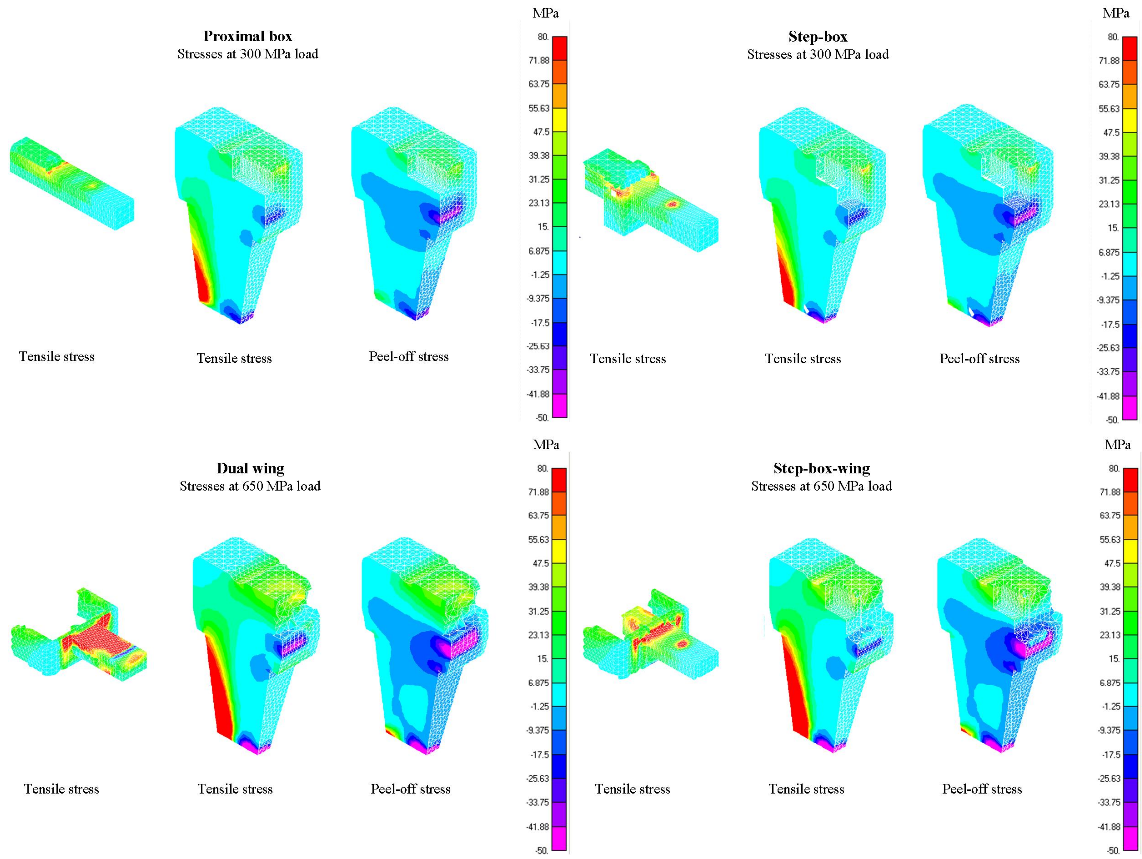

Figure 5.5 Distribution of tensile stresses and peel-off stresses in the tooth and the FRC framework of two-unit cantilever resin-bonded FRC-FDPs for different retainer designs.

5.5 Discussion

The main purpose of a dental reconstruction is functionally restoring the dentition. To fulfil this requirement a restoration should be able to withstand biting forces during mastication. Regardless the wide range of bite forces measured, the dental community seemed to have reached a consensus on the amount of load a reconstruction should be able to endure, namely 500 N in the premolar area [33,34]. With fracture strengths up to 697 N, this study proved that only dual-wing-retained and step-box-wing-retained FRC-FDPs are able to withstand these biting forces and consequently usable in the premolar region. Inlay-retained FRC-FDPs, on the other hand, appeared to fail at significant lower fracture strengths than their dual-wing-retained and step-box-wing-retained counterparts, loads to failure far below 500 N makes them unsuitable for the replacement of a single premolar. Dyer *et al.* [35] acquired similar results with direct three-unit FRC-FDPs and showed that slot-retained FRC-FDPs failed at lower loads than wing-retained and slot-wing-retained FRC-FDPs. An increased bonding surface can be obtained when using wings, which results in higher bond strength values because of more efficient stress transfer to the abutment teeth and lower stresses at the adhesive interface.

Compared to three-unit fixed-fixed designs, it was expected to find lower fracture strength values for two-unit cantilever designs. These lower values could be expected as a fixed-fixed design is considered, based on the simple beam theory, to suffer a lower amount of stress than a cantilever design. Romeed *et al.* [36] confirmed this assumption by investigating the mechanical behaviour of a three-unit fixed-fixed FDP and a two-unit cantilever FDP with 2D FEA. Although, it should be noted that they did not included the cement layer in their study.

Only one study reports on three-unit fixed-fixed inlay retained FRC-FDPs with a framework made of Estenia C&B EG Fiber and veneered with Estenia C&B and obtained a slightly higher value of 943 (233) N [37]. The inter-abutment distance for this study corresponded to a molar replacement of 15 mm, which was double the distance of the premolar gap (7 mm) in our study. It was shown before that inter-abutment distance has influence on fracture strength of inlay-retained FRC-FDPs [38]. Özcan *et al.* [33] reported on fracture strength values for a three-unit fixed-fixed design of a premolar replacement with comparable pontic span. They found an average fracture strength value of 1161 (428) N for conventionally prepared three-unit inlay-retained FDPs made of an everStick-framework and veneered with Tetric Ceram.
Unlike no significant differences were found between both wing-retained designs, the suggestion can be made that step-box-wing-retained FDPs could be slightly stronger than dual-wing-retained FDPs. The difference in predominant mode of failure between both designs, tooth fracture within the step-box-wing-retained group versus adhesive and/or FDP failure within the dual-wing-retained group, and the results obtained by Dyer *et al.* [35] corroborates this assumption.

In this study the amount of fibres incorporated in the FRC framework and the dimensions of the connector differed between inlay-retained FDPs and wing-retained FDPs. Inlay-retained FDPs contained only one bundle of FRC because the lack of space. Two bundles of FRC were used for dual-wing-retained FDPs, where each wing contained one bundle of FRC. Also two bundles of FRC were used for step-box-wing-retained FDPs. In this design the inlay contained one bundle of FRC, while each wing contained half a bundle of FRC. The use of two bundles of FRC caused an increase in connector-size for wing-retained FRC-FDPs. Although, the fracture strength values of wing-retained FDPs were significantly higher than those of inlay-retained FDPs, fracture mode analysis suggests that the difference in connector size and fibre amount were not the factors that caused the increase in fracture strength. The FRC-FDPs never failed due to fracture of neither the connector nor the retainer. Nevertheless, an increase in fibre amount as well as of connector-size can have a beneficial effect on the strength of FRC-FDPs [27,28,39,40].

Recent *in vitro* research by Li *et al*. [27,28] revealed the beneficial effect of adjacent teeth on anterior cantilever resin-bonded FRC-FDPs. Higher fracture strength values were obtained in specimen with adjacent teeth [28]. The observed effect was more important for non-reinforced than for reinforced specimen, respectively 47% and 11%. This finding was in agreement with the results of a subsequently conducted FEA study were lower stresses occurred in a model with adjacent teeth [27]. Such set-up obviously resembles closer to clinical reality and suggests that an amount of occlusal loading can be transferred to the adjacent teeth. With this in mind, based on the fracture strength tests, a better clinical performance of two-unit cantilever resin-bonded FRC-FDPs could be expected. The high fracture strength obtained for wing-retained FRC-FDPs in this study and the fact that the beneficial effect of adjacent teeth is more important in non-reinforced bridges [27,28] are convincing results that two-unit wing-retained non-reinforced resin composite FDPs could be used for single tooth replacement in the premolar area.

The failure mode analysis revealed that inlay-retained and step-box-wing-retained FDPs predominantly failed because of tooth fracture, which demonstrates the weakening effect of intra-coronal restorations. Previous research on fracture resistance of intact, prepared and restored posterior teeth showed that tooth preparation and restorations, like inlays, not only weakens a tooth, but also makes them more prone to cusp fracture [41-43].

The failure modes of the four FDP designs could be explained by 3D FEA. FDPs with a proximal box retainer or a step-box retainer all failed due to tooth fracture. In these cases a part of the proximal wall on the left and the right of the box preparation together with the FDPs fractured out of the abutment tooth. FEA revealed that the highest tensile stresses, which are apparently of the same magnitude as the strength of the tooth material, are in the same area. The highest peel-off stress is apparently lower than the bond strength between the tooth and the retainer. Highest tensile stresses in step-box-wing-retained FDPs presented at the central groove of the occlusal surface, were tooth fracture started, which made this design more prone to catastrophic cusp fracture. Dual-wing-retained designs predominantly failed due to debonding, pontic fracture or a combination. In the FEA the wing-retained designs showed the lowest tensile stresses, which are apparently below the strength of the tooth material and the highest peel-off stresses of all four designs were found in the occlusal area of the proximal surface, which explains why they often debonded. Comparison of the stress distribution in all four FRC frameworks revealed that the wing-retained designs suffered the largest amount of stress, which was far below the flexural strength of EG Fiber. The large amount of stress in the FRC frameworks suggests that proper fibre-reinforcement and framework design is of utmost importance for wing-retained FDPs.

The elastic modulus of 39 GPa for the EG fibre, as provided by the manufacturer, is higher compared to the 25 GPa obtained from three-point flexure testing [31]. Nevertheless, an elastic modulus of 39 GPa seems correct, as a similar value, provided by a different manufacturer, is used by Magne *et al*. [44]. In the 3D FEA model the elastic modulus of the FRC was decreased from 39 GPa to 20 GPa. This resulted in an increase of the maximum solid major principle stress from 52.3 MPa to 59.8 MPa and an increase of the maximum solid S_x stress increased from 48.8 MPa to 56.3 MPa. So, fibre-reinforced composite with a lower elastic modulus results in higher stresses at the adhesive interface, as well as in the tooth. Lower fracture strengths and more adhesive failures can be expected. The same principle accounts for PFC-FDPs.

It should be noted that the FEA models have some limitations, *e.g*. a simplified tooth model only composed of dentine, and a rigid adhesive interface instead of an elastic resin luting cement interface. The FEA model was created for revealing the major stress distribution in order to explain failure mode. The highest tensile stresses (52.3 MPa – 70.0 MPa) in the tooth were in range with the ultimate strength of dentine found in literature, respectively 54 MPa when tubules were orientated parallel to the shear plane and 92 MPa when tubules were orientated perpendicular to the shear plane [45]. Highest peel-off stresses (40.5 MPa – 48.8 MPa) at the adhesive interface were slightly higher than the micro-tensile bond strength of Panavia F to enamel and dentine, respectively 38.8 MPa and 17.5 MPa, reported in literature [46]. Although, the tooth in our FEA model was composed only of dentine, the restorations in our specimen were mainly bonded to enamel. It must be noted that the micro-tensile bond strength values reported by Hikita *et al.* [47] were obtained with rectangular specimens who were trimmed to a cylindrical hourglass shape with a diameter of 1.2 mm at the biomaterial-tooth interface. It was determined by Phrukkanon *et al.* [48] that micro-tensile bond strength values obtained with cylindrical hourglass shaped specimen results in values who underestimate real bond strength due to stress concentration at the biomaterial-tooth interface. 3D FEA models showed that the highest peel-off stresses occurred at surfaces were the restorations were luted to enamel. Visual inspection of the fractured specimen revealed that adhesive failures mainly presented at the bond surface between enamel and resin luting cement (Figure 5.6). So, we can conclude that 3D FEA was able to explain the observed predominant failure modes.

The choice for a dual-wing retainer was based on the fact that such retainers are believed to transfer and subsequently bear forces designated from dynamic tooth contacts more effective than a one-wing retainer. Both wings were 4 mm in length in order to establish a 180-degree wraparound, which improved the retention and resistance of resin-bonded bridges [9]. However, future research should determine the need on tooth preparation in this large extend. The step-box-wing retainer was tested because small mesial and/or distal Class 2 restorations are frequently present in (pre)molars. Based on the results of this study, a dual wing is the preferred retainer for replacing a lost premolar by means of an indirect two-unit cantilever resin-bonded FRC-FDPs, the strength is comparable with the step-box-wing retainer and the dominant mode of failure is debonding instead of cusp fracture. For these reasons we advise not to incorporate existing restorations into an indirect two-unit cantilever resin-bonded FRC-FDPs. Future research is needed to confirm this hypothesis. In such cases we propose the following procedure. To start with, the tooth should be restored with a direct resin composite restoration suitable for use in the posterior area. Tooth preparation and impression taking can be done immediately proceeding restoration at the same visit or during the course of a second visit. The dual-wing-retained FRC-FDPs should be placed, under dental dam isolation, in a last visit.

The limitations of this study must be recognised. The fact that the specimens were not subjected to artificial aging, such as thermo cycling and/or mechanical loading should be seen as a drawback. Static fracture strength testing after artificial aging resembles the clinical reality closer than without artificial aging. Also, this study is limited to high static loads, while in clinical conditions dental reconstructions are also exposed to low cyclic loading or fatigue loading. Failure of dental restorations is quite often caused by fatigue loading, in that respect future *in vitro* research should focus on fatigue. It is difficult to correlate *in vitro* tests in general and fracture strength and fatigue tests in particular to clinical reality. Therefore the authors recommend evaluation of this treatment modality during a proper designed randomized clinical trial before introduction as general dental practice.

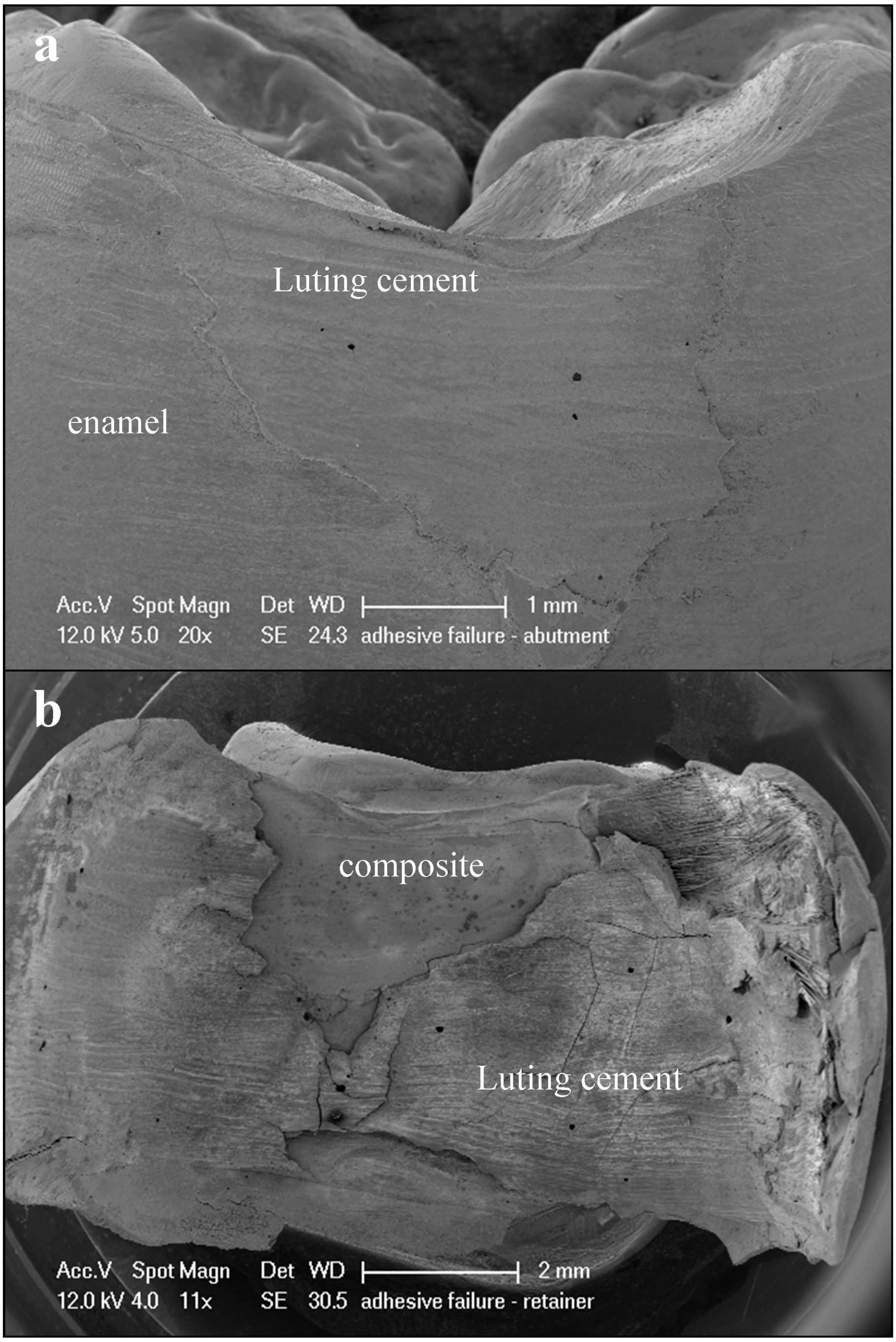

Figure 5.6 SEM micrographs of an adhesively failed dual wing retained FRC-FDP: (a) tooth interface; (b) retainer interface.

5.6 Conclusions

Within the limitations of this study it was concluded that a dual wing retainer is the optimal design for replacement of a single premolar by means of a two-unit cantilever resin-bonded FRC-FDP. The strength is comparable with the step-box-wing-retained FDPs while the predominant failure is debonding instead of catastrophic cusp fracture, which is more favourable.

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CHAPTER 6

The influence of framework design on the load-bearing capacity of laboratory-made inlay-retained fibre-reinforced composite fixed dental prostheses

6.1 Abstract

Objectives: Delamination of the veneering composite is frequently encountered with fibre-reinforced composite (FRC) fixed dental prosthesis (FDPs). The aim of this study is to evaluate the influence of framework design on the load-bearing capacity of laboratory-made three-unit inlay-retained FRC-FDPs.

Materials and Methods: Inlay-retained FRC-FDPs replacing a lower first molar were constructed. Seven framework designs were evaluated: PFC, made of particulate filler composite (PFC) without fibre-reinforcement; FRC1, one bundle of unidirectional FRC; FRC2, two bundles of unidirectional FRC; FRC3, two bundles of unidirectional FRC covered by two pieces of short unidirectional FRC placed perpendicular to the main framework; SFRC1, two bundles of unidirectional FRC covered by new experimental short random-orientated FRC (S-FRC) and veneered with 1.5 mm of PFC; SFRC2, completely made of S-FRC; SFRC3, two bundles of unidirectional FRC covered by S-FRC. Load-bearing capacity was determined for two loading conditions ($n = 6$): central fossa loading and buccal cusp loading.

Results: FRC-FDPs with a modified framework design made of unidirectional FRC and S-FRC exhibited a significant higher load-bearing capacity ($p < 0.05$) (927 ± 74N) than FRC-FDPs with a conventional framework design (609 ± 119N) and PFC-FDPs (702 ± 86N). Central fossa loading allowed significant higher load-bearing capacities than buccal cusp loading. This study revealed that all S-FRC frameworks exhibited comparable or higher load-bearing capacity in comparison to an already established improved framework design.

Conclusions: S-FRC seems to be a viable material for improving the framework of FRC-FDPs. Highest load-bearing capacity was observed with FRC frameworks made of a combination of unidirectional FRC and S-FRC.

6.2 Introduction

A fixed dental prosthesis (FDP) is considered as treatment of choice for replacing missing teeth. Since conventional and implant-retained FDPs are invasive, time-consuming, and expensive the dental profession continues the search for alternatives. One such alternative is a fibre-reinforced composite fixed dental prosthesis (FRC-FDP). FRC-FDPs are basically made of a fibre-reinforced composite framework acting as a stress dissipater and are veneered with particulate filler composite.

Following the introduction of glass fibre-reinforced composites in the early 1990s [1] their use increased enormously over the last years [2]. Limited information is available on their longevity and clinical behaviour, but the available clinical research showed that FRC-FDPs are able to function acceptably for up to five years [3-6], with reported 5 year-survival rates between 73% [5] and 93% [4]

Regardless of the promising results typical kinds of failures, like delaminating and chipping of veneering composite, were encountered during clinical function [3,5-7]. To overcome these failures, the framework design should be modified to support the veneering composite and the amount of fibres should be increased to improve the rigidity of the FPDs [6]. The most frequently used FRC framework consists of a bundle of unidirectional FRC placed in the central part of a FDP (Figure 1B). It seems that the amount of FRC included in such conventional framework is too little to provide the necessary support and rigidity. A high-volume anatomically-shaped FRC framework should be able to deal with these shortcomings.

Already some evidence, *in vitro* as well as *in vivo*, is available in the literature on framework design of FRC-FDPs. Behr *et al*. [8] tested simulated three-unit FRC-FDPs with one anatomical framework and two conventional framework designs and obtained significant higher fracture resistance for an anatomically-shaped framework (902 N) in comparison to conventional frameworks (694 N and 737 N). Also Xie *et al*. [9] tested the fracture resistance of inlay-retained FRC-FDPs with different framework designs. A framework which supported the pontic area in buccolingual direction showed significant higher fracture resistance compared to conventional and high-volume designs.

Freilich *et al*. [6] evaluated the clinical performance of short-span FRC-FDPs and changed during the course of the study the framework design. The original low-volume framework design, suffered veneer fractures in an early stage. Therefore a high-volume design, which was more rigid and offered more support for the veneering composite, was introduced. The high-volume design showed a 95% survival rate instead 62% for the low-volume design after a mean observation time of 3.75 years. Monaco *et al*. [7] investigated the clinical behaviour of inlay-retained FRC-FDPs with conventional and modified framework designs over a period of 12 to 48 months. The conventional framework design showed a higher failure rate than the modified framework design. In the group of FDPs with a conventional framework design delamination occurred in three cases (16%), while in the modified frame work group only one FDP (5%) suffered from chipping.

Short glass-fibres containing fibre-reinforced composite (S-FRC) with semi-interpenetrating polymer network matrix was recently introduced to dentistry [10]. Random-orientated S-FRC exhibit isotropic properties in comparison to the anisotropic properties of unidirectional fibres. S-FRC exhibit improved mechanical properties with regard to flexural strength and toughness in comparison to PFC [10,11]. Both properties make S-FRC a possible alternative to easily fabricate a high-volume anatomically-shaped FRC framework. Garoushi *et al*. [12] showed that short span FRC-FDPs made of S-FRC exhibited similar load-bearing capacity as conventional FRC-FDPs.

The aim of the present study was to evaluate *in vitro* the influence of framework design on the load-bearing capacity of laboratory-made inlay-retained FRC-FDPs. The null-hypothesis to be tested was that incorporation of S-FRC to FRC frameworks of FRC-FDPs improves their load-bearing capacity and generates a more favourable fracture pattern.

6.3 Materials and Methods

Eighty-four laboratory-made three-unit inlay-retained FRC-FDPs replacing a lower first molar were constructed. The FRC frameworks were made of a commercially available unidirectional E-glass-containing FRC (everStick C&B, Sticktech ltd, Turku, Finland) and a new experimental S-FRC. S-FRC was prepared as described previously [10]. The FRC frameworks were veneered with hybrid PFC for indirect use (Gradia-dentine A3, GC Corp., Tokyo, Japan). The materials used in this study and their composition are listed in Table 6.1.

Table 6.1 Materials used in this study.

Brand	Composition	Manufacturer	Lot number
Gradia Dentine A3	Resin: UDMA, EDMA; Filler: silica (≈ 75 vol%)	GC corp, Tokyo, Japan	0506021 0608221 0609111
everStick C&B	Resin: PMMA, Bis-GMA; Filler: silanised E-glass fibres (≈ 65 vol%)	Sticktech Ltd., Turku, Finland	2061010-ES-165
Experimental S-FRC	Resin: Bis-GMA, TEGDMA; Filler: silanised E-glass fibres (≈ 22.5 wt%), silanised silica particles (≈ 55 wt%)		
Multilink Sprint	Base paste: Resin: Bis-GMA, TEGDMA, UDMA; Fillers: barium glass, ytterbium trifluoride, silica; initiators/stabilizers Catalyst paste : Resin: Bis-GMA, TEGDMA, UDMA; methacrylated phosphoric acid ester; Fillers: barium glass, ytterbiumtrifluoride, silica; initiators/stabilizers	Ivoclar-Vivadent, Schaan, Liechtenstein	J22739

Bis-GMA bisphenol-A-gycidyl dimethacrylate; UDMA urethane dimethacrylate; EDMA ethylene dimethacrylate; UTMA urethane tetramethacrylate; PMMA poly(methyl methacrylate) Mw 220,000; TEGDMA triethylenglycoldimethacrylate.

FDP preparation

A zirconia model (Ice Zirconia, Zirconzahn, Bruneck, Italy) of a mandibular second premolar, a missing first molar and second molar, prepared to accommodate a three-unit inlay-retained FDP, was created (Figure 6.1). The inter-abutment distance of 11 mm corresponds with the mesial-distal dimensions of a mandibular first molar. The second premolar received a disto-occlusal inlay preparation (step: 3.0 x 2.0 mm; box: 1.5 x 3.5 mm; depth: 2.0 mm) and the second molar a mesio-occlusal inlay preparation (step: 4.0 x 3.0 mm; box: 1.5 x 5.0 mm; depth: 2.0 mm) according to the guidelines for composite inlay restorations. Preparations were made with conventional diamond burs (set 4278, Komet, Lemgo, Germany) in a water-cooled airrotor.

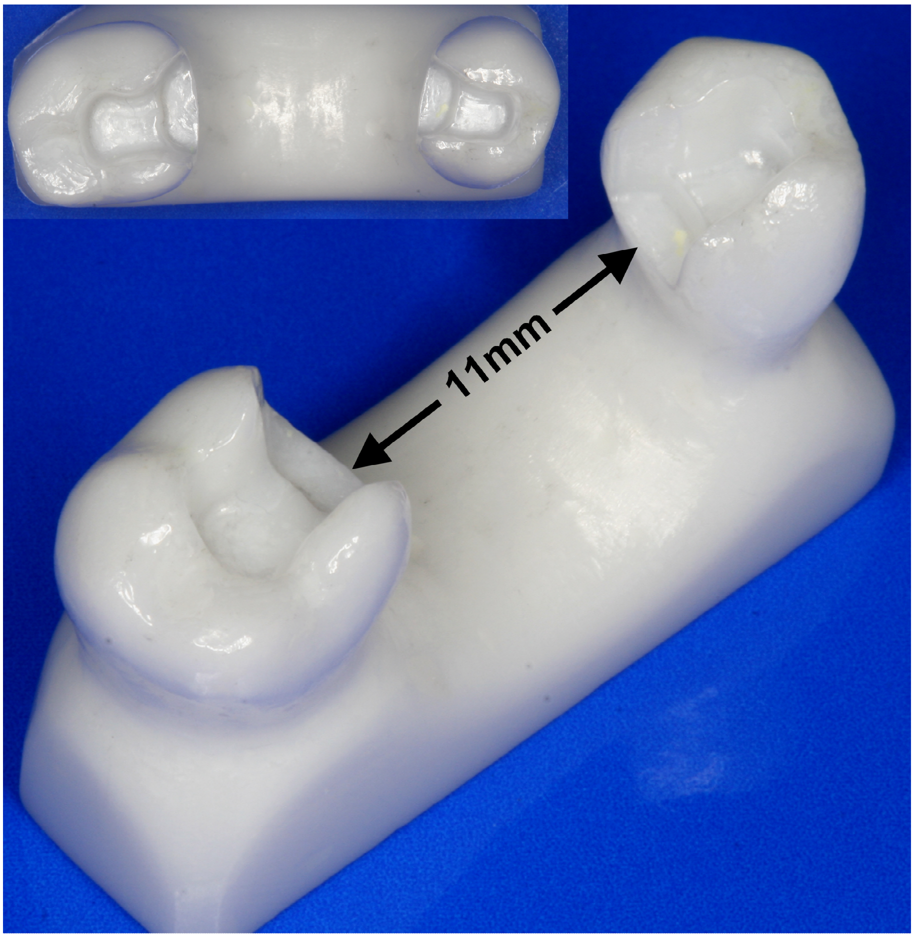

Figure 6.1 Test set-up used in this study: zirconia model representing a mandibular situation of a missing first molar. The second premolar and the second molar received two-surface inlay preparations in order to accommodate a three-unit inlay-retained FDP.

The FRC-FDPs were fabricated according to seven different framework designs (Figure 6.2):

PFC: made of PFC without fibre-reinforcement.

FRC1: made of PFC reinforced with one bundle of unidirectional FRC.

FRC2: made of PFC reinforced with two bundles of unidirectional FRC.

FRC3: made of PFC reinforced with two bundles of unidirectional FRC and two pieces placed perpendicular to the main framework.

SFRC1: made of an anatomically-shaped FRC framework, composed of two bundles of unidirectional FRC and experimental S-FRC, and veneered with 1.5mm of particulate filler composite.

SFRC2: made of experimental S-FRC.

SFRC3: made of experimental S-FRC and two bundles of unidirectional FRC.

FRC1 and FRC2 are conventional framework designs, while FRC3, SFRC1, SFRC2 and SFRC3 are modified framework designs.

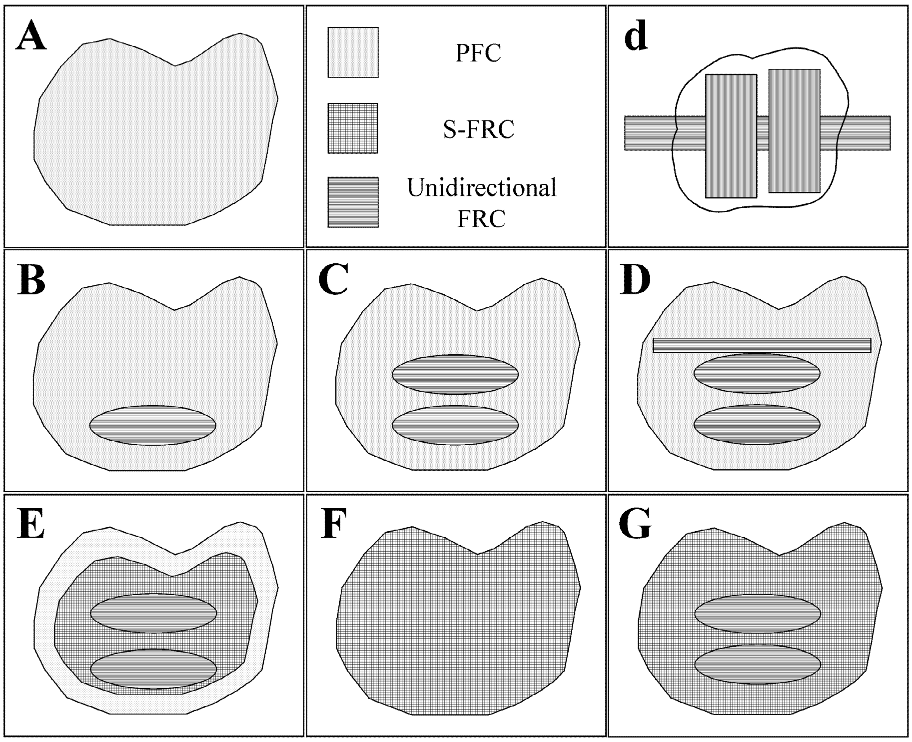

Figure 6.2 Graphical representation showing the cross-sections of the different framework designs used in this study. (A) PFC: PFC without fibre-reinforcement; (B) FRC1: PFC reinforced with one bundle of unidirectional FRC; (C) FRC2: PFC reinforced with two bundles of unidirectional FRC; (D) FRC3: PFC reinforced with two bundles of unidirectional FRC and two pieces placed perpendicular to the main framework; (d) FRC3: occlusal view; (E) SFRC1: anatomically-shaped FRC framework; (F) SFRC2: experimental S-FRC; and (G) SFRC3: experimental S-FRC and two bundles of unidirectional FRC.

The FRC framework was light cured for 10 s by a handheld polymerisation unit (Optilux 501, Kerr, CT, USA) with a power output of 800 $mW \cdot cm^{-2}$. The retainer and the molar pontic were veneered with hybrid PFC for indirect use (Gradia, GC Corp.). A transparent polyvinylsiloxane template (Memosil 2, Heraeus-Kulzer, Hanau, Germany) was used to standardise the dimensions and occlusal morphology of each FRC-FDP. Connector dimensions for the premolar were: height 4.0 mm; width 5.0 mm, and for the molar: height 4.5 mm; width 5.5 mm. Each increment was light cured for 20 s by the same handheld polymerisation unit. The completed FDP was post cured by light and heat in a light furnace (Lumamat 100, Ivoclar-Vivadent, Schaan, Liechtenstein) for 25 min. The specimens were dry stored for 24 h prior to luting.

The three-unit FDPs were luted to the zirconia model with a recently introduced self-adhesive, dual-curing resin luting cement (Multilink sprint, Ivoclar-Vivadent, Schaan, Liechtenstein). Pre-treatment of the adhesive surface of the inlay restorations was obtained by sandblasting (Cojet prep, 3M Espe, St Paul, MN, USA) with 30 μm silica-coated alumina particles (Cojet sand, 3M Espe) under 0.3 MPa pressure for 10 s followed by cleaning with compressed air for 5 s. No pre-treatment was required for the zirconia model. Excess luting cement was removed with a microbrush after the FDP was seated. Resin luting cement was light cured from three directions (occlusal, buccal, and lingual) for 40 s by a handheld polymerisation unit. The luted FDPs were left undisturbed for an additional 15 min to allow the resin luting cement to set.

Load-bearing capacity

Specimens were loaded until failure in a universal testing machine (model LRX, Lloyd instruments Ltd, Fareham, UK) at a crosshead speed of 1mm·min^{-1} and data were recorded using PC software (Nexygen, Lloyd instruments Ltd). The load was applied by a 6 mm diameter steel contact ball, as previously described [9,12]. Each group of FRC-FDPs was randomly divided into two subgroups (n = 6), which were subjected to two different loading conditions: for the first group the load was applied in the central fossa of the pontic (Figure 6.3A), while for the second group the load was applied to the buccal cusp (Figure 6.3B). The specimens were loaded till initial first signs of damage could be observed. Identification of initial failure was based on criteria described by Dyer *et al*. [13]: (1) a sharp decline in the load/displacement curve, (2) visible signs of fracture, (3) audible emissions, if at least two of the following conditions were present, initial failure was identified as such.

Fractured specimens were submerged in a methyl blue dye for 10min followed by 30s rinse with tap water. Specimens were visually examined and their mode of failure was recorded. Randomly selected specimen were sectioned (Isomet 1000, Buehler, Lake Bluff, IL, USA) in order to determine the cross-sectional FRC-volume.

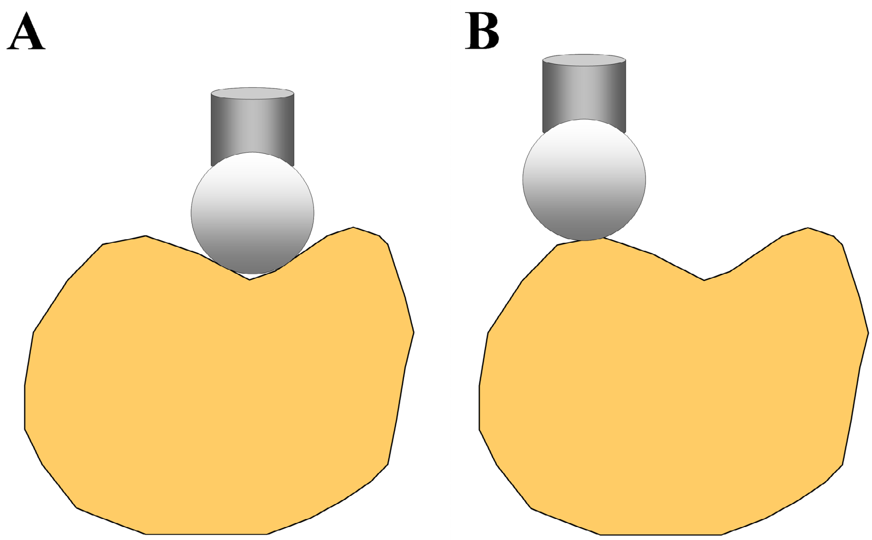

Figure 6.3 Graphical representation showing both loading conditions used in this study. (A) central fossa loading; (B) buccal cusp loading.

Statistical Analysis

Statistical analysis was performed with the statistical software SigmaStat 3.0 (SPSS Inc. Chicago, IL, USA). Mean and standard deviations of load-bearing capacities for each group were calculated. Two-way analysis of variance (ANOVA) followed by Tukey's post hoc test was performed to determine the effect of framework design and load condition on the observed load-bearing capacities. P-values of less than 0.05 were considered to be statistically significant.

6.4 Results

Load-bearing capacities (in N) of FRC-FDPs with different framework designs are graphically represented in Figure 6.4. Significant differences in load-bearing capacity were found between both loading conditions. Central fossa loading produced significant higher load-bearing capacities than buccal cusp loading for all groups ($p < 0.05$), except for FRC2. No strong differences between the different framework designs were revealed. Slightly higher load-bearing capacities were obtained for modified frameworks in comparison to conventional and PFC frameworks. Only SFRC3 (927 ± 74 N) was significant different from PFC (702 ± 86 N), FRC1 (609 ± 119 N), and FRC2 (592 ± 98 N) for central fossa loaded specimens. For buccal cusp loaded specimens, not only SFRC3 (751 ± 148N) was significant different from PFC

(403 ± 62 N), FRC1 (469 ± 80 N), FRC2 (483 ± 117 N), and FRC3 (529 ± 122 N), but also SFRC2 (643 ± 68 N) was significant different from PFC (403 ± 62 N).

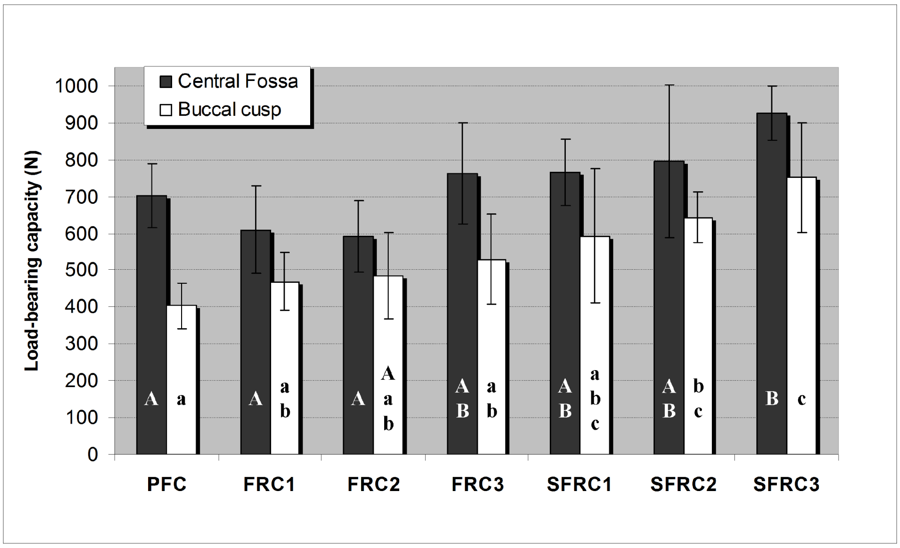

Figure 6.4 Load-bearing capacity of FRC-FDPs with different framework designs. Error bars showing the standard deviation. Groups denoted with the same superscript are not statistically different (Two-way ANOVA, Tukey multiple comparison, $p < 0.05$).

Visual inspection revealed three different failure modes: cracks, delamination and pontic fractures. Modes of failure for the different groups are shown in Table 6.2. Catastrophic failures were only seen for PFC when loaded at the central fossa. FRC1 and FRC2 suffered from delamination in up to 50% of the cases. Also one delamination failure occurred in FRC3 when loaded in the central fossa. Cracks were the most common failures and their location was uniform throughout the groups. The cracks originated from the gingival part of the connector towards the loading point (Figure 6.5).

Table 6.2 Fracture patterns of FRC-FDPs with different framework design.

Fracture pattern	PFC		FRC1		FRC2		FRC3		SFRC1		SFRC2		SFRC3	
	CF	BC	CF	BC	CF	BC	CF	BC	CF	BC	CF	BC	CF	BC
Cracks	0	6	3	5	3	4	5	6	6	6	6	6	6	6
Delamination	0	0	3	1	3	2	1	0	0	0	0	0	0	0
Pontic fracture	6	0	0	0	0	0	0	0	0	0	0	0	0	0

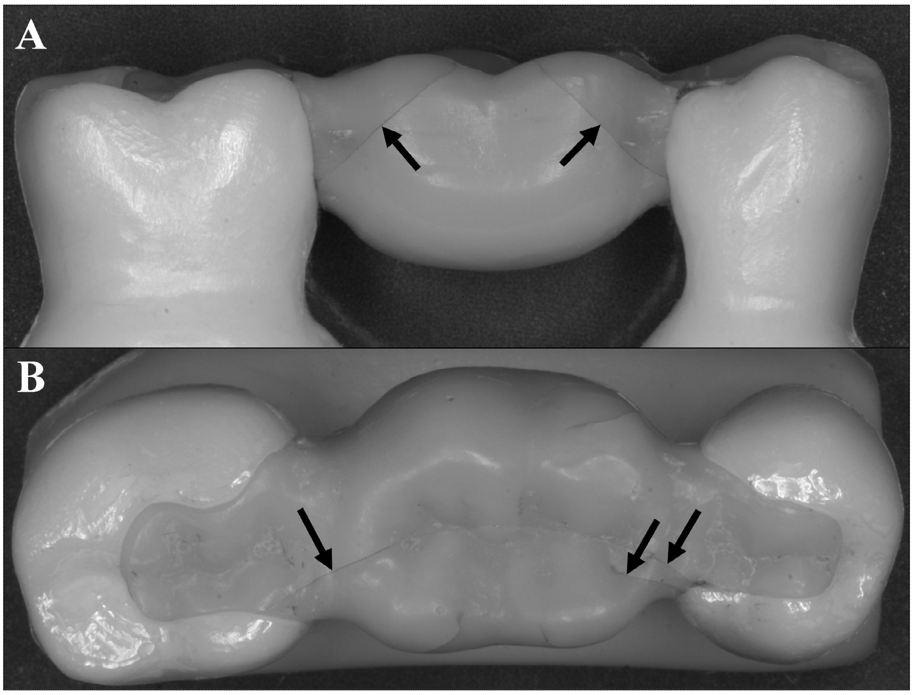

Figure 6.5 Failed FRC-FDP (group FRC2) showing cracks (black arrows) originating from the gingival part of the connector towards the loading point.

6.5 Discussion

Dental reconstructions are during clinical function subjected to biting and chewing forces. Functional rehabilitation of the dentition is the main purpose of a dental prosthesis. For that reason a FRC-FDP should be capable to withstand up to 500 N in the premolar region and 500 to 900 N in the molar region [14,15]. Previous research stated that FRC-FDPs are capable of bearing posterior biting forces [9,12,13,15,16]. Taking important aspects as initial failure and buccal loading into consideration suggests that FRC-FDPs with a conventional design and even some with a modified design (FRC3 and SFRC1) are maybe not indicated for use in the molar region. Nevertheless, it should be taken into consideration that the rigidity of the used test set-up negatively influences the values obtained in this study and underestimate the load-bearing capacity and subsequent clinical performance of FRC-FDPs. Load-bearing capacity values obtained in this study are situated in the lower range of those reported in literature. Previously reported load-bearing capacity values of FRC-FDPs range from 524 N [14] till 2500 N [9]. This wide range of values can be explained by the differences in study design: used materials, pontic span, retainer preparation and test set-up.

Although promising results were found during clinical studies, delamination of the veneering composite was frequently observed. In order to overcome those problems it was proposed to improve the FRC framework in a way it becomes more rigid and gives more support to the veneering composite, which was confirmed by several studies [6-9]. Increased rigidity of FRC frameworks can easily be obtained by increasing the amount of fibres. No significant difference was found between FRC1 and FRC2 indicating that increased framework rigidity alone seems insufficient. To increase the supportive nature of a FRC framework it should be constructed in such a way that the veneering composite can be uniformly supported. The modified FRC frameworks tend to produce slightly higher, but not always significant different, load-bearing capacities than PFC-FDP and conventional FRC frameworks (Figure 6.4). A previous study by Dyer *et al*. [13] indicated that significant differences between reinforced and unreinforced groups occurred only above a cross-sectional FRC-volume of 43%. Analysis of the pontic cross-sections of this study pointed out that the cross-sectional FRC-volume was far below 43% for all groups except SFRC2 and SFRC3, 4.8% and 31% respectively. Surprisingly, FDPs made of PFC showed a slightly higher load-bearing capacity, when loaded at the central fossa, than FDPs with a conventional FRC framework. This observation is in agreement with earlier findings by Dyer *et al*. [13] revealing that load-bearing capacity tends to be lower for low-volume FRC-FDPs in comparison to PFC-FDPs. This effect was observed for initial failure, but not for final failure. The load-bearing capacities values obtained in this study were also initial failure values. It has to be noticed that a distinguished difference with regards to failure pattern was found between PFC-FDPs and the other groups. PFC-FDPs suffered from catastrophic pontic failure, while FRC-FDPs suffered from delamination and veneer cracks. For that reason one should be aware of the fact that initial and final failure is the same for PFC-FDPs. When analysing the modified FRC frameworks it is noticed that the use of S-FRC slightly improves the performance of FRC-FDPs in comparison to an already established design (FRC3) [9]. The veneered S-FRC framework (SFRC1) showed to be slightly more supportive than FRC3 when loaded at the buccal cusp. It should be noted that evaluation of the cross-sectional design revealed a discrepancy between the ideal (Figure 6.6B) and the experimental design (Figure 6.6A), which can partially be explained by the unfavourable handling properties of S-FRC. From a clinical point of view one should be aware that such a design seems difficult to fabricate and proper training of dentist and dental technician is paramount. It can be hypothesized that an ideal design as depicted in Figure 6.6B would produce higher load-bearing capacity values more closely to SFRC2 and SFRC3. These results showed that FRC frameworks fabricated of S-FRC produced the highest load-bearing capacity values and will probably show the least chipping and delamination during clinical function. Nevertheless, it has to be noticed that the use of non-veneered S-FRC is associated with some important drawbacks, e.g. watersorption, aesthetics, polishability, and handling, which restricts its clinical use. For that reason groups SFRC2 and SFRC3 are not yet suitable for clinical application.

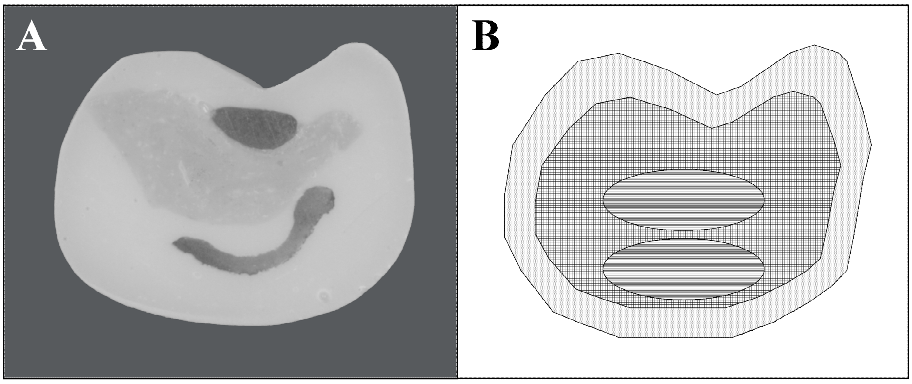

Figure 6.6 Representation of the discrepancy between (A) the obtained and (B) the ideal cross-section of FRC-FDP with an anatomic framework design (SFRC1).

Analysis of the failure patterns of FRC-FDPs pointed out that only PFC-FDPs encountered catastrophic failure presented as pontic fracture when loaded at the central fossa. Buccal cusp loading, on the other hand, only produced cracks, which can be attributed to the more complex stress pattern generated by the applied loading. The failure pattern of conventional framework designs not only presented as cracks, but also as delamination, the latter proving the insufficient support provided by these framework designs. Failure of modified framework designs presented as cracks indicating increased rigidity and supportive nature of these designs. The one delamination that occurred in FRC3 can be attributed to less careful framework construction. Closer inspection of the particular specimen revealed that the perpendicular placed fibre bundles were too short, which compromised the support of the cusps. Although, the increase in load-bearing capacity between conventional and modified framework designs was limited, failure analysis corroborates the improved performance of modified framework designs.

Central fossa loading is the most common used loading condition in static fracture strength testing of FDPs. In this study FRC-FDPs were loaded in the central fossa or at the buccal cusp of the pontic. Higher load-bearing capacities observed for fossa loading in comparison to cusp loading, which was in agreement with the results of of Xie *et al*. [9], confirms the latter to be far more demanding. This can be partially be explained by the fact that the fibre is loaded during fossa loading while the much weaker composite is loaded during cusp loading. A second explanation deals with the type of stresses induced by each loading condition. Fossa loading subjects FDPs to compressive stresses located beneath the loading point, tensile stresses located in the gingival part of the pontic as well as on the occlusal part of the connector and shear stresses located in the connector area. Cusp loading induces additional torsion stresses in the connector area and shear stresses in the cusps of the pontic. Those shear stresses in the pontic area are able to provoke chipping and delamination of the veneering composite.

The rationale for recording initial failure above final failure was based on previous research [13,15]. The mechanical performance of FRC-FDPs is overestimated when ultimate strength or final failure load values are considered. One should be aware of the fact that final failure loads can be 27% to 46% higher than initial failure loads [13,15]. It was stated by Dyer *et al*. [13] that it may be more valuable to search for reinforcement and designs that elevates the initial failure load of FDPs instead of the final failure load. The damage that arises at initial failure loads presented, in this study, as cracks or delaminations. This damage weakens the FDP and may initiate further degradation. Cracks act as easy and fast access points enabling oral fluids to penetrate the FRC. Semi-IPN matrix-based FRC more prone to watersorption in comparison to UTMA matrix-based FRC [17] or PFC [10], which can be explained by the filler content [10] and hydrophilic properties of the resin matrix [18]. Watersorption induces plasticisation of the resin matrix and deteriorates the fibre-polymer interphase by possible leaching of glass forming oxides from the fibre surface and by hydrolytic degradation of the polysiloxane network formed after silanisation of the glass fibres [18,19]. The above described mechanisms affect the mechanical properties of FRC resulting in lower strength and elastic modulus, the latter contributes to decreased rigidity of the framework.

Rigidity of the used test set-up could have influenced the load-bearing capacities in a negative way. Fischer *et al*. [20] showed that the fracture load of FDPs with rigidly mounted abutments decreased with 13% in comparison to non-rigidly mounted abutments. Additional bending stresses are induced in FDPs which are mounted in a rigid test set-up [20]. Not only could the rigidity of the test set-up, but also the elastic modulus of the abutments have had an influence on the load-bearing capacities. Non-rigidly mounted abutments with an elastic modulus close to that of natural teeth are capable of giving a more realistic representation of the oral situation. Such a set-up will generate a more evenly distributed stress pattern and subsequently generate higher load-bearing capacities.

Several studies showed that modified framework designs perform better under static loading conditions. Further research should focus on the fatigue behaviour of these modified framework designs.

6.6 Conclusions

Within the limitations of this study, the following conclusions can be drawn:

1. All framework designs exhibit higher failure loads when loaded at the central fossa than at the buccal cusp.
2. S-FRC improves the load-bearing capacity of FRC-FPDs.
3. Modified framework designs suffered less delamination than conventional designs.

This study revealed that all S-FRC modified frameworks exhibited comparable or higher load-bearing capacity in comparison to an already established modified framework design. So S-FRC seems to be a viable material for improving the framework of FRC-FPDs. Highest load-bearing capacity were observed with FRC frameworks made of a combination of unidirectional FRC and S-FRC.

6.7 References

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CHAPTER 7

Summary and conclusions

The general introduction in **Chapter 1** gives background information on the various available solutions for replacement of a single tooth. A rationale for the use of resin-bonded fixed dental prostheses (RB-FDPs) and especially two-unit cantilever RB-FDPs is provided. Subsequently, the use of FRC for the manufacturing of fixed dental prostheses (FDPs) is described, while framework design and clinical performance of fibre-reinforced composite fixed dental prostheses (FRC-FDPs) is discussed in detail. The second part of the general introduction focuses on the material science behind fibre-reinforced composites. First of all resin-based composites in general and fibre-reinforced composites in particular are introduced. The classification of resin-based composites depending on their general composition is explained. Subsequently, the composition of dental fibre-reinforced composites is discussed in detail. Thermosetting polymers, such as dimethacrylates and epoxies, are identified as the most widely used matrix polymers, while the most popular fibre reinforcement is glass fibre. The manufacturing of dental fibre-reinforced composite (FRC) is described. Furthermore, the principles of fibre reinforcement, the flexure- and fatigue properties of FRC are discussed. The aim of this thesis is to provide a better understanding of the influence of fibre-reinforcement on mechanical properties and prosthesis designs.

The first part of the thesis focuses on the mechanical properties of particulate filler composites (PFC) and FRC and bridges the gap towards their potential use in cantilever resin-bonded fixed dental prostheses (RB-FDPs).

Previous studies reported a beneficial effect of fibre-reinforcement on the fracture strength and fatigue resistance of dental resin-based composites, but never took the multi-vectorial nature of the forces occurring during physiological function into account. In **chapter 2** the influence of fibre-reinforcement on the fracture strength and fatigue resistance of resin-based composites was investigated. The fracture strength was obtained by means of a cantilever beam test, while the fatigue resistance was obtained using a rotational cantilever beam fatigue testing device. The rotational fatigue testing method exerts a multi-vectorial stress on the specimens to be tested in a sequence of tension and compression each cycle. In fact, the direction of the applied stress represents the clinical situation where stresses on occlusal surfaces vary from parallel to the surface to perpendicular. The used test methodology subjects each point of the circumference of a specimen repeatedly to tensile stress. It was concluded that fibre reinforcement has a beneficial effect on fracture strength, fatigue resistance, work-of-fracture, and failure mechanism of resin-based composites. The high fatigue resistance and their favourable failure mechanism make FRCs useful in stress bearing situations.

The first step in closing the gap between material properties and clinical behaviour is to design an *in vitro* test which provides information on how the material responds to clinical circumstances. Beams made of PFC and FRC luted to bovine enamel simulated two-unit cantilever RB-FDPs in **chapter 3**. This test set-up made it possible to investigate the influence of fibre-reinforcement and luting cement on the static and dynamic failure load of simulated two-unit cantilever RB-FDPs. The static failure load was determined with a peel test, which was identified by earlier research as most clinical relevant test. The dynamic failure load was once again determined with a rotating cantilever beam fatigue test. This study pointed out that fibre reinforcement has a significant effect on static and dynamic failure load of simulated two-unit cantilever RB-FDPs, which was not influenced by the type of luting cement. This study revealed also a difference in failure behaviour between PFC beams and FRC beams. PFC beams fractured, leaving the bonded part on the tooth surface, while FRC beams partially debonded from the tooth surface, leaving fibres connected to the enamel surface. It should be taken into consideration that the fibre reinforcement fulfils a fail-safe situation, because even after connector fracture the fibre reinforcement protects the FDP from complete debonding. Coincidentally, uncured FRC turned out to be prone to aging after their packaging was opened several months previously. A significant drop in dynamic failure load was observed for aged FRCs. It was concluded that FRC seems suitable for the fabrication of two-unit cantilever RB-FDPs, but the question how FRC performs in comparison to other materials, such as metals and all-ceramics remains.

The second series of studies focussed on prosthesis design. Initially, the use of FRC for designing anterior as well as posterior cantilever RB-FDPs was evaluated. Both studies can be seen as the second step in bridging the gap towards clinical reality.

Three dimensional finite element analysis was used to study the mechanical behaviour of anterior two-unit RB-FDPs made of different framework materials in **chapter 4**. The model consisted of a two-unit cantilever RB-FDP replacing a maxillary lateral incisor with a wing-shaped retainer on the central incisor. Five different framework materials such as, direct fibre-reinforced composite, laboratory fibre-reinforced composite, metal, glass-ceramic, and zirconia were compared. It was concluded that RBFDP made of FRC provided a more evenly distributed stress pattern from the loading area towards the abutment tooth than the other framework materials. Maximum principal stress was identified at the occlusal embrasure of the connector for all framework materials, which highlights the importance of proper connector design. Advanced stress analyses suggested a difference in predominant failure mode; connector fracture for FRC-, and glass ceramic-based RB-FDPs and debonding for metal-, and zirconia-based RB-FDPs. A stress concentration was found at the contact area between the pontic and the adjacent tooth, indicating that a part of the applied load is transferred towards the adjacent tooth. Such favourable stress transfer should be recognised by clinicians and researchers as an important design factor potentially influencing longevity of FRC-FDPs.

During the course of the previous study the question rose if cantilever FRC-FDP would be eligible for use in the posterior area of the oral cavity. Therefore, in **chapter 5** the influence of retainer design on the strength of two-unit cantilever glass fibre-reinforced RB-FDPs was investigated. Two inlay retained designs and two wing retainer designs were evaluated for replacing a missing premolar with a dummy attached to a first molar. The finite element analysis approach was used to reveal the stress distribution in order to be able to explain the observed failure modes. The wing-retained RB-FDPs showed significant higher fracture strengths than the inlay-retained RB-FDPs. A dual wing with 180 degrees wrap around was due to its favourable failure mode identified as the ideal retainer design for replacement of a single premolar with a two-unit cantilever glass fibre-reinforced RB-FDP.

Chipping and delamination of the veneering composite, due to inadequate design of the FRC framework, is identified as one of the most frequently occurring failures with FRC-FDPs under clinical conditions. Therefore, in **chapter 6** the influence of framework design on the load-bearing capacity of laboratory-made three-unit inlay-retained FRC-FDPs for loading in the occlusal fossa and at the buccal cusp was evaluated. A new anatomical framework design made of unidirectional FRC and short glass-fibre containing fibre-reinforced composite (S-FRC) was proposed and compared towards a non-reinforced design, two conventional designs and three modified framework designs. First of all, this study revealed that the load-bearing capacity of FRC-FDPs was affected by the loading condition. Clinicians should be aware of the fact that FRC-FDPs are more prone to failure during eccentric movements, which highlights the need of cusp protected or a well-balanced occlusion in combination with FRC-FDPs. It was concluded that S-FRC improves the load-bearing capacity of FRC-FPDs and that modified framework designs suffered less delamination than conventional designs. Therefore, delamination and chipping of the veneering composite can be reduced by using an anatomical framework design made of unidirectional FRC and S-FRC.

The series of studies conducted in this thesis showed that FRC is superior in comparison to PFC with regard to strength and fatigue properties. Furthermore, these investigations proved the potential of FRC to be used in an indication that can be regarded as the worst case scenario, namely two-unit cantilever RB-FDPs. Even posterior cantilever RB-FDPs seem within reach. In conclusion it was shown that a recently introduced S-FRC is a viable material for designing a new anatomical framework. The introduction of new indications and new prosthesis designs into daily clinical practice can not be based on laboratory studies only. Therefore, future research should focus on the evaluation of these new developments during properly designed randomized clinical trials.

Samenvatting en conclusies

In de introductie van dit proefschrift, **hoofdstuk 1**, wordt achtergrondinformatie gegeven aangaande de diverse mogelijkheden die er bestaan voor solitaire tandvervanging. De reden voor het gebruik van adhesiefbruggen en vooral éénvleugelige cantilever adhesiefbruggen wordt gemotiveerd. Aansluitend wordt het gebruik van vezelversterkte composiet voor de vervaardiging van vast brugwerk beschreven en werd dieper ingegaan op de voorhanden zijnde literatuur omtrent het ontwerp van de vezelversterkte onderstructuur en het klinisch gedrag van vezelversterkt brugwerk. Het tweede deel van de introductie richt zich op de materiaalkundige aspecten van vezelversterkte composieten. Allereerst worden tandheelkundige composieten in het algemeen en vezelversterkte composieten in het bijzonder geïntroduceerd. De classificatie van tandheelkundige composieten uitgaande van hun samenstelling wordt toegelicht. Vervolgens word uitvoerig ingegaan op de samenstelling van tandheelkundige vezelversterkte composieten. Na een grondige literatuurstudie werden thermohardende polymeren, zoals dimethacrylaten en epoxies, geïdentificeerd als de meest toegepaste matrix polymeren, terwijl glasvezel het vaakst werd gebruikt als vezelversterking. Het productieproces van tandheelkundige vezelversterkte composieten is beschreven. Ter afsluiting zijn de principes van vezelversterking en de buig- en vermoeiïngseigenschappen van vezelversterkte composieten besproken. Het doel van dit proefschrift is meer inzicht te verwerven aangaande de invloed van vezelversterking op enerrzijds de mechanische eigenschappen van tandheelkundige composieten en anderzijds op het ontwerp van adhesiefbrugwerk.

Het eerste deel van dit proefschrift richt zich op de mechanische eigenschappen van deeltjesgevulde composieten en vezelversterkte composieten en tracht een brug te slaan naar hun toepasbaarheid bij het vervaardigen van éénvleugelige cantilever adhesiefbruggen.

Ondanks dat voorgaande studies reeds over het gunstig effect van vezelversterking op de breuksterkte en vermoeiïngsweerstand van tandheelkundige composieten rapporteerden, werd tot op heden nooit echt rekening gehouden met het multivectorieel karakter van de krachten die tijdens fysiologische functie plaatsgrijpen. In **hoofdstuk 2** wordt de invloed van vezelversterking op de breuksterkte en vermoeiïngsweerstand van tandheelkundige composieten onderzocht. De breuksterkte is bepaald doormiddel van een cantilever buigtest, terwijl de vermoeiïngsweerstand geregistreerd is doormiddel van een rotatie vermoeiïngstest. De rotatie vermoeiïngstest oefent een multivectoriële kracht uit, die varieert tussen trek- en drukkracht, op elk te testen specimen. Door de richting van de uitgeoefende kracht te variëren wordt de klinische situatie, waar de kauwkrachten zowel parallel als loodrecht op het occlusaal oppervlak aangrijpen, beter benaderd. De gebruikte testmethode onderwerpt elk omtrekspunt van een specimen herhaaldelijk aan trekspanningen. Deze eerste studie concludeerde dat vezelversterking een gunstig effect heeft op zowel breuksterkte, vermoeiïngsweerstand en breuktaaiheid, alsook op het faalgedrag van tandheelkundige composieten. In het bijzonder vezelversterkte composieten zijn omwille van hun hoge vermoeiïngsweerstand en gunstig faalgedrag bijzonder aangewezen in situaties waar grote krachten verwacht worden.

Het ontwerpen van een laboratorium test die informatie geeft over hoe het materiaal reageert in klinische omstandigheden vormt de eerste stap in het proces om te trachten een brug te slaan tussen enerzijds materiaaleigenschappen en anderzijds klinisch gedrag. Om éénvleugelige cantilever adhesiefbruggen na te bootsen werden, in **hoofdstuk 3**, platte balkvormige specimen gemaakt van vezelversterkte en deeltjesgevuld composiet die werden gecementeerd aan het glazuur van rundertanden. Deze test opstelling maakte het mogelijk om het effect van zowel de vezelversterking als het bevestigingscement op de statische en dynamische afbreukwaarden van gesimuleerde éénvleugelige adhesiefbruggen te testen. De statische afbreukwaarden werden bepaald met een peeltest, welke door eerder onderzoek geïdentificeerd werd als de meest klinische relevante test. De dynamische afbreukwaarden werden opnieuw bepaald met een rotatie vermoeiïngstest. Deze studie toonde aan dat de statische en dynamische afbreukwaarden van gesimuleerde éénvleugelige cantilever adhesiefbruggen voorzien van een vezelversterking significant verhoogde, doch geen effect ondervonden ten aanzien van het gebruikte bevestigingscement. Deze studie bracht eveneens een verschil in faalgedrag naar voor tussen gesimuleerde éénvleugelige cantilever adhesiefbruggen gemaakt van deeltjesgevuld en vezelversterkte composiet. Specimen gemaakt van deeltjesgevuld composiet braken doormidden, waardoor het gecementeerde deel achterbleef op het tandoppervlak. vezelversterkte specimen kwamen deels los van het tandoppervlak, waardoor een beperkte hoeveelheid vezels aan het glazuur bleef kleven. Men moet zich terdege bewust zijn van het feit dat een vezelversterking fungeert als een soort veiligheidsriem, die een brugwerk na breuk van het connectordeel beschermt tegen loslaten. Bij toeval werd vastgesteld dat niet gepolymeriseerd vezelversterkte composiet verouderde, nadat hun verpakking enkele maanden eerder geopend werd. De dynamische afbreukwaarden van deze composieten bleken aanzienlijk lager te liggen dan deze van de vezelversterkte composiet uit niet geopende verpakkingen. Deze studie concludeerde dat vezelversterkte composieten geschikt lijken voor de fabricage van éénvleugelige cantilever adhesiefbruggen, maar de vraag blijft hoe vezelversterkte composieten presteren in vergelijking met andere materialen, zoals metalen en keramieken.

De tweede reeks onderzoeken van dit proefschrift bestudeerde het ontwerp van vezelversterkt brugwerk. In eerste instantie werd de mogelijkheid van het gebruik van vezelversterkte composiet voor het vervaardigen van zowel anterieure als posterieure éénvleugelige cantilever adhesiefbruggen geëvalueerd. Beide studies trachten ons weerom een stap dichter bij klinische realiteit te brengen.

In **hoofdstuk 4**.werd het biomechanisch gedrag van anterieure éénvleugelige cantilever adhesiefbruggen gemaakt van verschillende materialen bestudeerd aan de hand van een drie-dimensionale eindige elementen analyse. Het eindig elementen model bestond uit een éénvleugelige cantilever adhesiefbrug ter vervanging van een maxillaire laterale snijtanden met een retentievleugel ter hoogte van de centrale snijtanden. Vijf verschillende materialen werden met elkaar vergeleken nl. vezelversterkte composiet voor directe toepassing, vezelversterkte composiet gebruikt voor toepassing in het tandtechnisch labo, metaallegering, glaskeramiek en zirkonia. Dit onderzoek toont dat de via eindig elementen analyse berekende interne spanningsconcentraties het meest gelijkmatig verdeeld zijn in het geval dat éénvleugelige cantilever adhesiefbruggen vervaardigd worden van vezelversterkte composiet. Bij alle toegepaste materialen werd de hoogste spanningsconcentratie waargenomen in het connectordeel van de adhesiefbrug, meer bepaald ter hoogte van het occlusale deel van de insnoering tussen pontic en pijlerrestauratie. Deze spanningsconcentratie benadrukt het belang van een doordacht ontworpen connectordeel. Grondige analyse van de eindig elementen modellen laat zien dat er een verschil in faalgedrag verwacht kan worden afhankelijk van het gebruikte materiaal, nl. connectorbreuk in het geval van vezelversterkte composiet en glaskeramiek in tegenstelling tot loslaten in het geval van een metaallegering en zirconia. Tevens werd er een spanningsconcentratie aangetroffen op het contactvlak tussen pontic en buurtand, wat aangeeft dat een deel van de kauwkrachten overgebracht wordt naar de buurtand. Het belang van deze krachtoverdracht dient door clinici en onderzoekers herkend te worden, aangezien het in belangrijke mate de overleving van vezelversterkt brugwerk gunstig kan beïnvloeden.

Het vorige onderzoek riep de vraag op of vezelversterkte éénvleugelige cantilever adhesiefbruggen in aanmerking zouden komen voor toepassing in het posterieure deel van de mond. Om een antwoord op deze vraag te geven werd in **hoofdstuk 5** onderzocht welk type retentievleugel in deze situatie de hoogste breuksterkte leverde. In dit onderzoek werd een premolaar vervangen door een tweedelige cantilever adhesiefbrug te cementeren aan een eerste molaar, waarbij twee types intracoronaire restauraties en twee types retentievleugels toegepast werden als pijlerrestauraties. Het faalgedrag van de verschillende restauraties werd in tweede instantie verklaart na analyse van het intern spanningspatroon door middel van eindige elementen analyse. Er werden significant hoger breuksterktes geregistreerd voor cantilever adhesiefbruggen met een retentievleugel als pijlerrestauratie dan voor deze met een intracoronaire restauratie. Mede op basis van het gunstig faalgedrag werd de mesiale retentievleugel die het element 180 graden omvat weerhouden als de ideale pijlerrestauratie voor een glasvezelversterkte tweedelige cantilever adhesiefbrug ter vervanging van een premolaar.

Een vaak voorkomend klinisch probleem bij vezelversterkt brugwerk is chipping of delaminatie van het composiet dat de vezelversterkte onderstructuur bedekt. Een vaak aangehaalde reden voor dit probleem is het slechte ontwerp van deze onderliggende vezelstructuur. Daarom werd in **hoofdstuk 6** geëvalueerd welke invloed het ontwerp van de vezelversterkte onderstructuur had op de breuksterkte van in het tandtechnisch labo vervaardigde driedelige vezelversterkte bruggen. De vezelversterkte bruggen werden op twee manieren belast, nl. ter hoogte van de occlusale fissuur en de buccale knobbel. Er werd een nieuw vezelversterkte onderstructuur met een anatomische vormgeving, vervaardigd uit een combinatie van unidirectioneel vezelversterkt composiet en composiet versterkt met korte glasvezels (S-FRC) voorgesteld. Dit nieuwe ontwerp werd vergeleken met een niet-vezelversterkte, twee conventionele vezelversterkte en drie geöptimaliseerde vezelversterkte onderstructuren. Allereerst bevestigde dit onderzoek dat breuksterkte van vezelversterkte bruggen beïnvloed wordt door het type belasting dat ze ondergaan. Tandartsen moeten zich ervan bewust zijn dat vezelversterkte bruggen vaker falen ten gevolge van articulatie beweging, waardoor een hoektandgeleiding danwel een gebalanceerde articulatie met dit type bruggen van primordiaal belang is. Ter conclusie kan men stellen dat een S-FRC in staat is de breuksterke van vezelversterkt brugwerk te verhogen, doch werd eveneens minder delaminatie van het overliggende composiet vastgesteld met de geöptimaliseerde in vergelijken met de conventionele vezelversterkte onderstructuren. Bij vezelversterkt brugwerk kan delaminatie en chipping van het overliggende composiet wel degeljk beperkt worden door een anatomisch vezelversterkte onderstructuur gemaakt van unidirectioneel vezelversterkt composiet en S-FRC toe te passen.

Dit proefschrift toont aan dat de sterkte en de vermoeiïngseigenschappen van vezelversterkte composiet superieur is in vergelijking met deeltjesgevuld composiet. Bovendien bleek het potentieel van vezelversterkt composiet om gebruikt te worden in een situatie die de ‘worst case scenario’ representeerd, namelijk éénvleugelige cantilever adhesiefbruggen. Dit type adhesiefbruggen lijkt eveneens in de premolaarregio binnen de mogeijkheden te liggen. Tot slot werd aangetoond dat een recent geïntroduceerd vezelversterkt composiet met korte vezels een mogelijke toepassing heeft voor het construeren van een vezelversterkte draagstructuur bij brugwerk. Het introduceren van nieuwe indicaties en nieuwe prothetische ontwerpen in de dagelijkse klinische praktijk kan niet alleen worden gebaseerd op laboratoriumonderzoek, daarom dient toekomstig onderzoek gericht op de evaluatie van deze nieuwe ontwikkelingen binnen het kader van degelijk opgezette gerandomiseerde klinische studies.

DANKWOORD

Nooit had ik van tevoren kunnen inschatten wat de impact van promotieonderzoek op je leven is. Zonder overdrijven kan ik zeggen dat dit, tot op heden, de meest intense periode in mijn bestaan geweest is. De ervaring die ik op zowel professioneel als persoonlijk vlak heb meegekregen is van onschatbare waarde. Na vijf jaar ACTA heb ik mijn iets of wat ‘conservatieve’, of noem het belgische, visie op de tandheelkunde aardig moeten bijstellen. Dat er op persoonlijk vlak heel wat plaatsgevonden heeft is wel het minste wat je kan zeggen. Eerst en vooral ben ik in het huwelijksbootje gestapt met een pracht van een vrouw en nadien ben ik nog eens vader geworden van een schattig dochtertje. Maar, ik ben ook mezelf flink tegen gekomen. Uiteindelijk moet ik dat positief bekijken, want ik weet hierdoor hoe mijn toekomst op professioneel vlak er moet uitzien. De mogelijkheid die ik gekregen heb om tijdens mijn promotieperiode naar Finland te gaan heeft waarschijlijk nog het meeste invloed op mij gehad en is misschien op zich al een proefschrift waard. Vandaag kan ik stellen dat een promotietraject heel wat meer inhoud dan testjes uitvoeren! Velen hebben zowel direct als indirect een bijdrage geleverd tot de voltooiing van dit proefschrift. Zonder iemand bewust te willen vergeten wil ik toch enkele mensen in het bijzonder bedanken:

Prof. dr. A.J. Feilzer, beste Albert, promotieonderzoek doen onder jouw vleugels heb ik ervaren als een voorrecht. Één woord van jou was meestal voldoende om het onderzoek een grote sprong voorwaarts te laten maken. Jouw visie op tandheelkunde en research heeft me vaak aan het denken gezet en heeft mijn wetenschappelijk denken uiteindelijk in belangrijke mate vorm gegeven. Maar ook jouw menselijke kant heeft een diepe indruk bij mij nagelaten. Jij voelt je medewerkers perfect aan, waardoor je in staat bent om ze op een heel bijzondere manier te motiveren.

Dr. C.J. Kleverlaan, beste Cees, wat zouden we doen zonder jou? De inzet die jij toont voor de vakgroep en het onderzoek is met geen woorden te beschrijven. De momenten dat ik weer eens bij je te raden kwam met een of ander probleem of vraag waren talrijk, maar steeds was jij in staat om een oplossing te voorschijn te toveren. De vele suggesties en verbeteringen die je aanbracht in de manuscripten, zorgden er steeds weer voor dat het ook daadwerkelijk artikels werden. Waar ik veel respect voor heb is de manier hoe je, als chemicus zijnde, je plaats gevonden hebt tussen al die ‘einzelgängers’ oftewel tandartsen.

Dr. A van Dalen, beste Anne, jouw eerste stappen in het glasvezelonderzoek vormde een draagvlak voor mijn promotieproject. Ik mag dan ook met trots stellen dat jij mee aan de wieg stond van dit proefschrift. Jouw kritische opmerkingen en taalkundige correcties vormde een belangrijke bijdrage aan de manuscripten. Uiteindelijk bleken er niet enkel overeenkomsten tussen ons terug te vinden op gebied van onderzoek, maar ook op persoonlijk vlak. Ik moet je dan ook bedanken voor de steun die je me gaf tijdens de moeilijkere periodes van mijn promotietraject.

Prof. dr. P.K. Vallittu, dear Pekka, at the beginning of my PhD project I never could have dreamt to be able to work with you, the father of glass fibre-reinforcement. Thank you for the opportunity you gave me to stay for 3 months at your department. Your view on fibre research and the research group you created is unique and inspired me a lot.

Prof. dr. M.A.J van Waas, beste Rien, we leerde elkaar kennen via de Dental Triangle bijeenkomsten toen ik nog in Brussel werkte. Geleidelijk aan groeide mijn interesse om naar ACTA te komen en te promoveren. Jij hebt het onderwerp in eerste instantie bij mij aangebracht en vond het verstandig om het onderzoek onder Albert's supervisie te laten uitvoeren, wat uiteindelijk een schot in de roos was. Ik moet je dan ook bedanken voor de kans die je me gegeven hebt en de interesse die je steeds bent blijven tonen voor mijn onderzoek.

De leden van de promotiecommissie, prof. dr. P.K. Vallittu, prof. dr. R.J.G. De Moor, prof. dr F.J.M Roeters en prof. dr. D. Wismeijer, wil ik bedanken voor de tijd en aandacht die ze hebben besteed aan het doornemen van dit proefschrift.

Paranimfen Lippo Lassila en Akikazu Shinya.
Lippo and Aki, my sauna buddies, we first met during my trip to Turku in November 2006. Our friendship has grown ever since. Today, I can state that both of you became inner circle friends.
Dear Lippo, it is not easy to find the right words to thank you, because your friendship strongly influenced me in many ways. Not only scientifically, but also personally. It was you who introduced me into the secrets of FRC research, Finland, the Finnish way of life, salmiakkikoskenkorva, Vappu, Juhannus, whisky and sauna. The last one is bit of an understatement, because it was more than just sauna, it was 'avantosauna'. Your biggest merit or crime is making me addicted to avantouinti or ice hole swimming. By

the way, as a paranymph you are not allowed to drink whisky during the defence of a thesis, but afterwards...
Dear Aki, we spend so many great moments together during my consecutive stays in Turku. It was always a pleasure to go for a beer at one of the ‘Old …’ pubs or for a sauna. Also remember the stressful but hilarious moments at the biomaterials lab and later at the TCBC. I really appreciated your visit to Brussels when Lise-Marie was born. Don’t worry, I’m not forgotten that I still owe you one more thing: visiting Japan. Now this PhD is finished we should make work of this promise.

Beste Joris, jouw vriendschap heb ik altijd weten te appreciëren. Ging het nu over onderzoek of over iets persoonlijk, het maakte niet uit, want bij jou kon ik steeds terecht voor een goed gesprek Bedankt voor de lekkere diners en de gezellige whiskymomenten. Het wordt de hoogste tijd om dit nog eens in België te organiseren, want je leven zal binnenkort heel wat hectischer worden met een baby erbij.

Arie Werner en Jacqueline Rezende, beste Arie en Jacqueline, julie moet ik bedanken voor de hulp bij het uitvoeren van de experimenetn en bij het bedienen van de testapparatuur. Arie, jouw ‘oerhollandse’ mentaliteit heb ik met de jaren leren appreciëren en zal ik uiteindelijk ook missen.

Mijn collega’s op de afdeling tandheelkundige materiaalwetenschappen wil ik danken voor de stimulerende werkomgeving en de leuke discussies aan de koffietafel: Niek, Pallav, Toon, Moustafa, Hessam, Hang, Alma, Leontine, Ghazal, Ana, Houda, Chenfeng, Tjalling, Peter, Rien, Maria, Tony,...

I can’t forget to thank all my colleagues and the people I’v met at the Institute of Dentistry in Turku (Finland) for their warm welcome and friendship: Makiha, Sufyan, Jasmina, Arzu, Anne, Riina, Helena, Ami, Mervi, Jukka, Jenni, Anne-Maria, Bora, Isil, Emre, Kohji, Tomo, Teemu, Sari, Tiina, Minttu, Sevi, Hana, Ahmed, Aous...

Dit is eveneens het uitgelezen moment om de studenten te bedanken die in het kader van hun wetenschappelijke stage een bijdrage leverde aan het onderzoek: Reinier en Jacqueline, Dana en Luke, Jeroen en Martijn.

I wish to thank the dental manufacturers, Sticktech Ltd., Kuraray Dental Benelux BV, 3M Espe and, GC Europe, for generously supplying us with the materials. Especially

Eija Säilynoja from Sticktech Ltd. and Joost Nederkoorn from Kuraray Dental Benelux BV are acknowledged for the constructive collaboration.

Mijn schoonbroer, Filip, wil ik bedanken voor het omslagontwerp en het begeleiden van het drukwerk.

Papa en mama, wat jullie voor mij gedaan hebben, heb ik onbewust steeds als een evidentie gezien. Ik besef maar al te goed dat dit niet zo is en dat jullie steeds meer deden dan eigenlijk nodig was. Met het voltooien van dit proefschrift wordt het dan ook de hoogste tijd om jullie in de bloemetjes te zetten, want het is jullie verdienste dat ik uiteindelijk zover gekomen ben. Bedankt voor alles.

Liefste Charlotte, de voorbije jaren waren niet altijd even eenvoudig. Pendelen tussen Nederland en België vergde vaak heel wat planning, waardoor er vooral op zondag vaak weinig tijd voor ontspanning was. Ik besef ook dat ik er vaak niet was op momenten dat je mijn steun en troost kon gebruiken. Desondanks ben je steeds mijn steun en toeverlaat gebleven. Ik besef maar al te goed wat jij voor mij betekent, want zonder jou aan mijn zijde zou ik niet zijn wie ik nu ben. Met het beeindigen van dit proefschrift komt er terug tijd vrij die aan jou toebehoord. Tijd die we samen kunnen besteden aan onze volgende projecten, nl. ons huisje bouwen en een zusje of broertje voor Lise-marie op de wereld zetten.

Allerliefste Lise-Marie, mijn kleine prinses, jij bent zonder enige twijfel het mooiste cadeau dat papa ooit gekregen heeft. Telkens ik naar Nederland vertrok moest ik met pijn in het hart afscheid van je nemen. Nu papa's 'boekje' klaar is hoeft dat niet meer. En wie weet, misschien komt er ooit een tijd dat ook jij een boekje gaat schrijven...

Bedankt!

CURRICULUM VITAE

Filip Keulemans werd geboren op 16 december 1975 te Mechelen (België). Hij volgde de richting wetenschappen-wiskunde en behaalde het ASO diploma in 1994 aan het Koninklijk Atheneum 2 te Mechelen. In hetzelfde jaar begon hij zijn studie tandheelkunde aan de Vrije Universiteit Brussel (VUB), waar hij in 2000 afstudeerde als tandarts. Tijdens zijn studie groeide zijn academische intresse, wat resulteerde in een deeltijdse baan als assistent conserverende tandheelkunde en kroon-en brugwerk aan de VUB en een deeltijdse baan als tandarts in een privépraktijk. Hij volgde van 2001 tot 2003 een postacademisch opleiding esthetische tandheelkunde aan de VUB. Geleidelijk aan groeide de interesse om te promoveren, wat in 2005 leidde tot een aanstelling als promovendus bij de afdeling tandheelkundige materiaalwetenschappen aan het Academisch Centrum Tandheelkunde Amsterdam (ACTA). In het kader van zijn onderzoek naar vezelversterkte composieten verbleef hij enkele maanden op de afdeling biomaterialen van de Universiteit van Turku in Finland. Filip is getrouwd met Charlotte Stolte sinds 2006. Hun dochter Lise-Marie werd geboren in april 2008 en momenteel zijn zij in blijde verwachting van een broertje of een zusje voor haar.

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